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Bionics limb prostheses: Advances in clinical and prosthetic care

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Bionics limb prostheses: Advances in clinical and prosthetic care

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Editorial: Bionics limb prostheses: Advances in clinical and prosthetic care

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Editorial on the Research Topic
Bionic limb prostheses: advances in clinical and prosthetic care

by Frossard L, Conforto S and Aszmann OC. (2022). Front. Rehabilit. Sci. 3:950481. doi: 10.3389/fresc.2022.950481

Context

Importance of residuum health

The use of a prosthesis is essential to maintain function and wellbeing of individuals suffering from limb absence (1, 2). Consequently, providers of prosthetic care recommend bespoke interventions to sustain lenient interactions between individuals' residual limb and their prosthesis (3–7). The clinical management of this interface is critical because it greatly affects the residuum health (8).

Residuum health is influenced by intrinsic determinants inherent to personal demographics (e.g., gender, age, weight, and height) and surgical amputation (e.g., length of bone, muscle reassignments, muscle strength, and adipose tissue distribution) and extrinsic determinant-associated attachment (e.g., socket design and direct skeletal attachment) and prosthetic components (e.g., choice and alignment of components, control of the prosthetic joint movements, use of walking aids, and level of activity) (8). In all cases, interactions between intrinsic and extrinsic determinants are critical as residual tissues have limited physiological capacity to withstand direct loading applied by typical socket-suspended prostheses during daily activities (e.g., chafing and rubbing) (5, 9–12). In addition to general neurological residuum and phantom pain, individuals can experience a range of incapacitating neuromusculoskeletal dysfunctions compromising residuum health, such as acute and chronic skin issues, edema, neuroma, tendinitis, muscle contractures, stress fractures, osteopenia, and heterotopic bone growth, which altogether increases the risks of sound lower joints osteoarthrosis, and hyperlordosis (6, 13, 14).

Consequently, satisfactory residuum–prosthesis interface is difficult to achieve and sustain (15). Individuals with compromised residuum health are more at risk to experience unsuccessful prosthetic fitting arrangements (4, 16). Those with healthy residuum are more likely to maximize comfort, stability, and mobility when using a

suitable prosthesis. Individuals tend to go up and down between low (e.g., bedridden, use of wheelchair, and two crutches without prosthesis), unsatisfactory (e.g., two crutches with prosthesis, one stick, and independent ambulation with pain), and satisfactory (independent ambulation without pain and participation in recreational and professional activities) levels of activity depending on their satisfaction with prosthetic fitting, functional abilities, and need for aids (17–20). Individuals are often trapped going back and forth between unsatisfactory and satisfactory health states depending on pain level with the prosthesis (19, 21). Pain leads to frequent, and too often permanent, prosthesis abandonment (22–24). Altogether, repeated episodes of care addressing prosthetic fitting generate great personal distress and heavy socioeconomic burdens (e.g., healthcare expenses and work absenteeism) (25–30).

Emergence of new bionic solutions

In the last few decades, we have witnessed promising developments in the production of bionic limb solutions that could possibly alleviate, separately or altogether, some of the residuum health and fitting issues (31–34). Some innovations provide better prosthetic attachment through osseointegrated implants that could either extend the residuum limb and facilitate socket fit (e.g., endoskeletal implant) or protrude the skin to allow the fitting of bone-anchored prostheses (e.g., endoskeletal-exoskeletal implant) (19, 35, 36). Other innovations aim predominantly to reduce pain and improve control of the prosthetic limbs, including regenerative peripheral nerve interfaces, targeted muscle reinnervation (TMR), agonist–antagonist myoneural interface, and sensory feedback (31, 37–40).

Altogether, these emerging bionic bone-anchored prostheses could dramatically alleviate socket-related issues and improve intuitive usage of artificial limbs (33, 41–43). Early evidence of the clinical outcomes of these new interventions has indicated that they have, altogether, the potential to engender life-changing benefits (e.g., body image, sitting comfort, osseoperception, pain reduction, prosthetic control, walking ability, and health-related quality of life) (44–47).

Need for more information about rehabilitation and prosthetic care bionic solutions

Reports of scientific advances of a particular solution tend to focus primarily on the design of interface between the body and the hardware (e.g., osseointegrated implants and electrodes), screening process (e.g., eligibility criteria), surgical techniques (e.g., number of stages and reinnervation matrices), fitting and design of prosthetic components (e.g., microprocessor-controlled joints and control algorithms) as well as short- to long-term outcomes (e.g., physical tasks and health-related quality of life) (48–52).

Although critical to successful clinical outcomes, rehabilitation procedures (e.g., training exercises) and prosthetic fitting recommendations (e.g., setting of components) for new solutions are often areas of continuous development and, therefore, are under-reported (53–59). The level of understanding and acceptance of pre- and postoperative clinical care may vary between interventions for lower or upper bionic limbs.

More information is needed to elucidate the relationships between surgical procedure, clinical care, prosthetic fitting, and outcomes of current and emerging interventions (e.g., efficacy and safety) that are critical for establishing an evidence-based reasonable, and eventually best, standard of care for current and future bionic solutions.

Contribution

Scope of the research topic

Initially, we identified a need for more information about:

- Preoperative interventions that could possibly maximize surgical and medical outcomes of bionic limb solutions (e.g., screening process, strength, and reconditioning, stretching program).
- Postoperative intervention following surgical insertion of osseointegrated implants (e.g., rehabilitation programs, prescription of loading progression, monitoring of loading exercises, design of static and dynamics load-bearing exercises, strength, and conditioning).
- Postoperative intervention after targeted muscle reinnervation (e.g., extraction of physiological signal, development of classifiers, design of fine and/or gross motor control training exercises, training for intuitive control).
- Fitting of bionic and/or bone-anchored prostheses (e.g., choice and alignment of prosthetic components, training with microprocessor-controlled joint units, fall prevention program).
- Short- and long-term outcome measures of efficacy and safety of bionic and/or bone-anchored prostheses extracted from standardized and non-standardized instruments (e.g., physical tasks and self-reported surveys).
- Quantitative evaluation of functional recovery with techniques based on kinematics and dynamics and on the processing of myoelectric signal of the non-amputee limb to study adaptation and recovery strategies also aimed at the optimal choice of prosthesis.

It will be unrealistic to expect that this Research Topic alone will outline the current "state-of-the-art" on these topics. Therefore, we decided to gather a series of highly focused articles presenting forthcoming ideas and concepts as well as preliminary data about current and emerging bionic solutions.

Outline of key contributions

This Research Topic features a total of 10 articles written by 54 authors (39% females and 61% males) from 23 institutions across 7 countries. It presents one Perspective, Review, Case Report, Brief Research Report, and six Original Research articles.

As detailed in **Table 1**, six manuscripts involved individuals with transtibial, transfemoral, hip disarticulation, and transhumeral amputations. Two other basic studies used cadavers and animal specimens. Six manuscripts focused on socket interface and three locked at the design of a percutaneous part, the osseointegration process, and the surgical procedure for direct skeletal attachment specific to bone-anchored prostheses. Four studies sought to improve safety of prosthetic care, more particularly reduction of fall, improvement of osseointegration, and reduction in the infection of future osseointegrated implants. Eight studies aimed at improving efficacy, particularly mobility and function, reduction of phantom and residuum pain, and control of prosthesis.

Overview of new bionic solutions

Raschke (2022) wrote an introductory review that provided critical insights into the historical developments of the prosthetic technology and practices within the greater context of successive industrial revolutions (Industry 1.0 to Industry 4.0). Raschke shared her astute perspective on the expected benefits of the current industry revolution. The unfolding Industry 4.0 is characterized by the convergence of physical, digital, and biological systems that support the creation of smart technology and cyber-physical systems enabling innovative bionic bone-anchored prostheses (e.g., advanced manufacturing, additive manufacturing, data analytics, augmented reality, simulation, horizontal/vertical integration, cybersecurity, cloud computing, and the industrial internet). Raschke also highlighted the importance of health economic assessments to determine the balance between the costs and the benefits of these innovations (25).

Taylor et al. (2022) used cadaveric mechanical testing, medical imaging, and finite-element analyses of humeri and tibia to improve the design of the percutaneous osseointegration docking system for direct skeletal prosthetic limb attachment. The translation of the exact system from the humerus to the tibia may not be suitable because of differences in impaction force and stress distribution. Each type of implant must be designed following a specific shape and mechanical constraints.

Bohart et al. (2022) used a porcine model to develop an infection-free integration between the skin and a percutaneous part of skin and bone integrated pylon for direct skeletal attachment of lower limb prostheses. Injections of botulinum toxin into the four thigh muscles of the distal thigh of the left hind leg were sufficient to provide noticeable immobilization the skin's movement around the implant by the fourth week after the procedure. Injections of botulinum toxin might limit skin movements around a percutaneous part of an implant, thereby possibly reducing postoperative risks of infection.

Borkowska et al. (2022) performed a randomized cross-over study within able-bodied participants to assess the capacity of a new haptic sleeve to improve mechanotactile feedback. This study locked at changes in weak, normal, and strong grasp using visual, haptic, or combined feedback. The mechanotactile feedback provided by the haptic sleeve effectively improve grasping tasks and reduced energy expenditure.

Bresslerf et al. (2022) asked clinicians and end users to complete a System Usability Scale survey and semistructured interview to validate a new computer-assisted limb assessment (CALA) tool that can standardize documentation and visualization of phantom limb sensations and pain and quantify the patient's body image. CALA allowed for an accurate description and quantitative documentation of phantom limb pain. This capacity to analyze, monitor, and report sensation and pain information can help to close the gap between the therapist's conception and the patient's perception of phantom limb sensation and pain.

Kannenberg et al. (2022) analyzed the outcomes of an online survey completed by 46 individuals with transtibial amputation to determine whether anecdotal reports on reduced musculoskeletal pain and improved patient-reported mobility were isolated occurrences or reflect a common experience in powered prosthetic ankle–foot users. Users reported improvements in mobility and reduction of sound knee and amputated side knee pain when using powered prosthetic ankle–foot compared with passive feet. However, a substantial proportion of powered prosthetic ankle–foot users also reverted to passive feet.

De Marchis et al. (2022) performed a multimodal prosthetic gait assessment using a series of kinematic, kinetic, and electrophysiological datasets collected on individuals with different types of amputations and prosthetic components for a project funded by the Italian Worker's Compensation Authority. This study showed the importance of analyzing movement neural control and mechanical actuation of prosthetic limb as a whole rather than through segregated analyses focusing specific aspects. Multimodal prosthetic gait

TABLE 1 Key descriptors of the studies presented in this Research Topic.

	Number of studies		Raschke (2022)	Taylor et al. (2022)	Bohart et al. (2022)	Borkowska et al. (2022)	Bresslerf et al. (2022)	Kannenberg et al. (2022)	De Marchis et al. (2022)	Finucane et al. (2022)	Bachini et al. (2022)	Boesendorfer et al. (2022)
	(#)	(%)										
Population												
Cadavers	**1**	**10**		×								
Animal	**1**	**10**			×							
Human	**7**	**70**				×	×	×	×	×	×	×
Able-bodied	**1**	**10**				×						
Amputees	**6**	**60**					×	×	×	×	×	×
Amputation												
All level	**1**	**10**					×					
Lower Limbs	**7**	**70**										
Transtibial	**3**	**30**		×				×	×			
Transfemoral	**3**	**30**							×	×	×	
Hip disarticulation	**1**	**10**								×		
Upper limb	**2**	**20**										
Transhumeral	**2**	**20**		×								×
Attachment												
Socket	**6**	**60**					×	×	×	×	×	×
Bone-anchored prosthesis	**2**	**20**		×	×							
Contribution												
Safety	**4**	**40**		×	×					×		
Fall	**1**	**10**								×		
Osseointegration	**2**	**20**		×	×							
Infection	**1**	**10**			×							
Efficacy	**8**	**80**	×			×	×	×	×	×	×	×
Mobility and function	**5**	**50**						×	×	×	×	×
Phantom pain	**2**	**20**					×				×	
Residuum pain	**1**	**10**						×				
Control	**1**	**10**				×						

The bold values represent the number of articles focusing on the key descriptors in the rows expressed in number of articles and percentage of whole articles (100% = 10 manuscripts published).

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assessment can facilitate a more effective design of prostheses and therapies for patients fitted with conventional and new bionic limbs.

Finucane et al. (2022) asked individuals with a unilateral transfemoral or knee disarticulation amputation to follow new training (i.e., verbal, visual, tactile cueing, and patient education) to improve functional mobility (i.e., level-ground walking, stair climbing, incline walking, and sit-to-stand transitions) with a powered knee and ankle prostheses. This study provided new training techniques that can help individuals fitted with lower limb prostheses to take advantage of these powered devices and achieve their desired clinical outcomes.

Bachini et al. (2022) asked an individual with transfemoral amputation to wear four prosthetic interfaces stimulating specific areas of the residual limb (e.g., rigid and a semirigid socket with and without a focal pressure) to investigate if socket design can influence phantom sensations. Phantom sensations were different during distinct phases of the walking gait cycle depending on the four interfaces and led to changes in some gait spatiotemporal parameters. Phantom sensations were modulated by the prosthetic interface and could provide natural somatosensory information dynamically varying with gait phases.

Boesendorfer et al. (2022) reported the experience and outcomes of an individual who opted for an elective arm amputation to solve the lack of function due to obstetric brachial plexus injury. The participant showed a distinct improvement of function and high wearing times of the prosthesis at follow-up assessment. Selected patients who experience severe neurological deficit of biologic hand function might benefit from the elective amputation and subsequent restoration with the bionic hand.

Next steps

Sparking discussions

As highlighted by Raschke (2022), the successful development of bionic solutions integrating physical, digital, and biological systems will occur through a multitude of small increments. This Research Topic contributes to this global effort as it identifies knowledge gaps while, hopefully, sparking discussions about these new concepts capable of advancing clinical and prosthetic care of bionic limb prostheses.

From concept to standard of care

These articles should motivate more teams to engage in formalized research and publications further advancing these innovations. Accumulation of evidence through registered clinical trials will be required to facilitate clinical adoption and subsequent acceptance as standard of prosthetic care. Robust evidence will be required to overcome what Harris (2016) described as the "decline effect" (e.g., Initial strong results of new treatments tend to fade overtime with subsequent independent and stronger studies) (60). This will be critical to convince public and private healthcare funding bodies to support a particular innovation, particularly with the emergence of the fee-for-device business model (e.g., hospital, work cover, and insurance) (25–30).

Toward a global ecosystem

These clinical and prosthetic care innovations will contribute to the formation of a global ecosystem where a set of organizations and services will integrate the value chain of these bionic solutions through various commercial models. This emerging ecosystem will include providers of prosthetic solutions and administrators of healthcare organizations. More importantly, consumers will be at the heart of the ecosystem through involvement in the co-design of innovations and influence of consumers' advocates. Involving all stakeholders will critical to warrant that these bionic innovations, indeed, improve safely the life of growing population of individuals suffering from limb loss worldwide.

Author contributions

LF contributed to conceptualization, methodology, investigation, data curation, writing—original draft, writing—review and editing, visualization, supervision, project administration, and funding acquisition. SC contributed to conceptualization, methodology, investigation, data curation, writing—original draft, writing—review and editing, visualization, supervision, project administration, and funding acquisition. OA contributed to conceptualization, methodology, investigation, data curation, writing—original draft, writing—review and editing, visualization, supervision, project administration, and funding acquisition. All authors contributed to the article and approved the submitted version.

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Conflict of interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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Finite Element Analysis of Transhumeral and Transtibial Percutaneous Osseointegrated Endoprosthesis Implantation

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Cadaveric mechanical testing of a percutaneous osseointegration docking system (PODS) for osseointegration (OI) prosthetic limb attachment revealed that translation of the exact system from the humerus to the tibia may not be suitable. The PODS, designed specifically for the humerus achieved 1.4–4.8 times greater mechanical stability in the humerus than in the tibia despite morphology that indicated translational feasibility. To better understand this discrepancy, finite element analyses (FEAs) modeled the implantation of the PODS into the bones. Models from cadaveric humeri ($n = 3$) and tibia ($n = 3$) were constructed from CT scans, and virtual implantation preparation of an array of endoprosthesis sizes that made contact with the endosteal surface but did not penetrate the outer cortex was performed. Final impaction of the endoprosthesis was simulated using a displacement ramp function to press the endoprosthesis model into the bone. Impaction force and maximum first principal (circumferential) stress were recorded to estimate stability and assess fracture risk of the system. We hypothesized that the humerus and tibia would have different optimal PODS sizing criteria that maximized impaction force and minimized first principal stress. The optimal sizing for the humerus corresponded to implantation instructions, whereas for the tibia optimal sizing was three times larger than the guidelines indicated. This FEA examination of impaction force and stress distribution lead us to believe that the same endoprosthesis strategy for the humerus is not suitable for the tibia because of thin medial and lateral cortices that compromise implantation.

Keywords: osseointegration, finite element, endoprosthesis, humerus, tibia

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INTRODUCTION

Percutaneous osseointegration endoprosthetic systems are surgically implanted into the medullary canal of amputated bone, and are then passed permanently through the skin and connected to distal exoprostheses. This process bypasses socket suspension, returning limb loading to the bone and proximal joints. Initial introduction was focused on transfemoral limb loss, with increasing utilization for transhumeral and transtibial amputations (1). The Percutaneous Osseointegration Prosthesis (POP) (DJO Surgical, Austin, TX, United States), for example, was developed by the Salt Lake City VA and University of Utah for transfemoral amputees. This system underwent

an extensive preclinical evaluation using animal models to refine design characteristics and implantation techniques (2–5). The POP was clinically introduced to 10 patients (Early Feasibility Study, ClinicalTrials.gov NCT02720159) who had improved 1-year post-operative functional outcomes, such as increased bone mineral density and decreased don/doff time (6, 7).

Subsequent transhumeral device development began for percutaneous osseointegration docking systems (PODSs) (8) mimicking the fixation strategy of a transfemoral device with a tapered porous-coated region, circular in cross-section. It is unknown if this same design approach is suitable for the tibia. A morphologic analysis of the tibia suggests that it is possible but may not be appropriate for all residual limb lengths (9).

Uniaxial mechanical testing of the PODS device on a cadaveric tibia assessed initial fixation of the device (**Supplementary Materials**). In this pilot study, PODS devices were implanted into the tibia and tested for torsion and axial pullout following the methods used for the humerus (8). The results revealed a large discrepancy between initial stability of the PODS system in the tibia compared to the humerus, where failure was 0.21 and 0.7 × that of the humerus in pullout (1,325.1 ± 185.8 N) and torsion (6 ± 2.6 Nm), respectively (8). Before proceeding with the PODS system for transtibial use, we must determine the mechanism for the decrease in initial stability, as this corresponds to the stability of the system early post-operatively when little bone OI has occurred and, worst-case scenario, when no OI occurs. Initial stability also serves as our current predictive measure of long-term stability while we do not have destructive mechanical testing results *in vivo*. Since the tibial morphology indicated a likely fit for the PODS system (9), evaluation of the mechanical interface between the bone and endoprosthesis of the humerus and tibia would determine mechanistically how the same OI region geometry of the endoprosthesis would have a different performance in the two bones. Two primary metrics that correlate to the initial stability of the bone-endoprosthesis interface were examined in this study: (1) impaction force and (2) first principal stress. Impaction force corresponds to the total traction force of the endoprosthesis and initial stability of the system (10). First principal stress corresponds to circumferential stress, which is the primary fracture modality for impaction testing of intramedullary endoprostheses and should be minimized to avoid failure (11–13).

Implantation of PODS devices currently depends on qualitative observation to properly size the residual bone to the endoprosthesis. Implantation instructions indicate that proper sizing is achieved when uniform cortical bone is removed around the distal circumference of the reamer (8). The same instructions were applied to the tibia during mechanical testing (**Supplementary Materials**). It is possible that the same size selection protocol used successfully for the humerus is not applicable for the tibia, because the medullary canal is less circular and does not have a uniform cortical thickness (9).

Finite element analyses were performed to evaluate percutaneous OI devices for transfemoral use during daily loading (14–17) and failure (18). These studies identified zones of stress shielding that could lead to bone resorption, and stress risers that could lead to bone or endoprosthesis failure. This approach has not been used to evaluate how an endoprosthesis design differs in initial stability for specific anatomic locations.

The goal of this study was to perform finite element analyses (FEAs) to understand the large difference in mechanical failure of PODS between the humerus and tibia. Impaction of the tapered PODS OI region was simulated in the humerus and tibia using cadaver-specific FEA models. Each was implanted with a range of endoprosthesis sizes, encompassing those that contacted the endosteal surface but did not penetrate the periosteal surface, to evaluate the influence of size on impaction force and circumferential stress. We hypothesized that the humerus and tibia would have a different PODS sizing criterion that optimized maximum impaction force and minimum circumferential stress.

METHODS

Finite Element Model Description

A total of three humeri and three tibiae were obtained. The use of cadaver tissue was deemed exempted by both the University of Utah Institutional Review Board and the Salt Lake City Veterans Affairs Medical Center (protocol #11755). No demographic information was available for the humeri (one left, two right), but measurements taken from CT scans showed that they were near average in length and cortical thickness (19). The tibiae (two left, one right) were from Caucasian male donors 18, 34, and 46 years old; 172-, 188-, and 183-cm tall; 86, 75, and 91 kg, respectively, and were near average in length and cortical thickness (9). These three representative bones from each anatomic location were selected to understand behaviors of the bone during implantation that correlate to the mechanical results of the same bone to elucidate the mechanism of mechanical failure in the tibia with this endoprosthesis geometry. The bones were scanned using a Siemens SOMATOM Definition Flash (Siemens) scanner (120 kVp, 100 mAs, 512 × 512 acquisition matrix, 1-mm slice thickness) with a bone density calibration phantom (qCT Pro Model 3 CT; Mindways Software Inc., Austin, TX, United States). The bones were then segmented and reconstructed in 3D (MIMICS v23.0; Materialise, Plymouth, MI, United States).

To compare the humerus and tibia more directly, the medullary diameter and average cortical thickness of the humerus at 30% amputation length, where mechanical testing was performed, were best matched to the medullary diameter and average cortical thickness of the tibia, resulting in a 40% amputation length (9, 19). The average diameter of the medullary canal for this distal osteotomy, before any additional bone preparation, was then recorded and corresponded to the indicated size of endoprosthesis according to surgical instructions provided by the manufacturer of the device (20). Each was then virtually reamed to replicate implantation procedures for PODS. A 6-cm tall, 2°-tapered endoprosthesis with a circular cross-section was subtracted from the bone reconstruction to match the prepared inner surface of the bone according to validated procedures for virtual implantation (20). The subtracted reamer was placed at the centroid of the

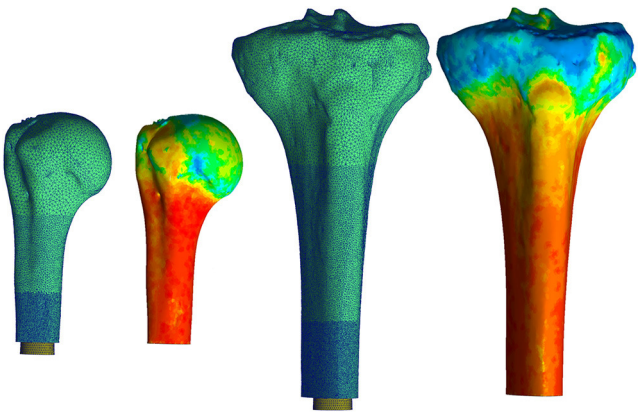

FIGURE 1 | Mesh for both the humerus (left) and tibia (right) began proximally with a coarse mesh controlled with a maximum edge length of 1.2 mm. This progressed into a finer mesh distally with a maximum edge length of 0.2 mm around the endoprosthesis. Heat map represents Young's Modulus values assigned to each element based on voxel intensity. Pictured humerus and tibia; Young's Modulus ranged from 0.007 (blue) to 14.8 GPa for the humerus and was 16.8 GPa for the tibia (red). The results of this material assignment were similar for all the other bones modeled. *Relative sizes of the bones are approximately to scale.

medullary canal and aligned coincidentally to the inertial axis of the medullary canal. The reamer diameter corresponded 1:1 to the endoprosthesis shape when the endoprosthesis was 3 mm proud the distal osteotomy, as measured from the resection to the distal collar of the endoprosthesis. This resulted in ∼0.1-mm radial interference between the bone and endoprosthesis when completely inserted, resulting in dilation of the bone (8, 20).

An endoprosthesis model was created as a 3-cm tall conical geometry with a circular cross section and 2° tapered angle (20). The smaller proximal diameter ranged from 8 to 20 mm, and referenced nominal endoprosthesis size. Each bone model was implanted with many endoprostheses that contacted the endosteal surface but did not penetrate the outer cortex. This allowed for a range of simulated sizes for each bone model to determine the impact of size selection.

Mesh geometry for all the models was constructed using commercially available software, (3-Matic; Materialise, Plymouth, MI, United States) and based on a convergence analysis of one representative tibia, as the region of interest is similar and the modeling approach is the same for both bones. Maximum element edge length dictated mesh parameters because of change in surface area due to different endoprosthesis sizes modeled. Distally, around the endoprosthesis contact zone and through the thickness to the periosteal surface, a maximum edge length of 0.2 mm was used. This grew proximally to a 0.7-mm maximum edge length between the contact zone and the surgical neck of the humerus and tibial tuberosity, followed by a 1.2-mm maximum edge length proximal to these anatomic landmarks (**Figure 1**). This mesh configuration yielded a 2% difference in impaction force and 5% difference in principal stress compared to a mesh refined to half of those maximum edge lengths. The chosen mesh ran in half the time (22 min, 42 s vs. 48 min 13 s using an Intel Core i5-7600 K processor with 16 GB RAM). The endoprosthesis model had a 0.5-mm maximum edge length across its entirety. All objects were assigned a four-node tetrahedral element. The four-node elements were selected because a 10-node tetrahedral element mesh resulted in only 3% change in force and 0.8% change in stress with 10× the amount of processing time. With small displacements, the use of a rigid body endoprosthesis, and by element-specific property assignment, the four-node tetrahedral elements were determined to be sufficient to represent these data.

Bone density was calculated with linear regression equations derived from the calibration phantom and applied to voxel intensity. These equations were dependent on a CT scanner and settings compared to known density values from the phantom. Young's modulus of the bone was assigned according to a study that performed regression analysis to correlate tibial, mid-diaphyseal, and cortical bone CT measurements to mechanical and physical properties (21):

$$E = 0.06^{*}\rho^{0.74}$$

All voxels ≥ 100 HU were equally divided into 10 uniform subgroups based on a convergence analysis; an increase to 20 subgroups resulted in only 3% decrease in maximum principal stress and 1% decrease in final impaction force. The median value for each subgroup became the assigned Young's modulus and density for each element in the subgroup (**Figure 1**). A lower-bound Young's modulus of 0.007 GPa and density of 0.05 g/cm^3 were assigned to all voxels ≤ 100 HU. No assumptions were made on the overall distribution of the material properties of the bone, since material property assignments of each element were made based on the voxel intensity of the CT scan compared to the calibration phantom in the field of view.

All amputated bone and endoprosthesis meshes were imported into FEBio Studio (v1.0, FEBio Software Suite, febio.org) (22). The bones were assigned neo-Hookean material properties but maintained the element-specific Young's modulus and density values and a uniform Poisson's ratio of 0.3 (17). It should be noted that FEBio automatically converts Young's modulus and Poisson's ratio to Lamé parameters, since this model is not solved as linear elastic and strains did not exceed infinitesimal strain assumptions. Since the endoprosthesis is made of titanium, which is much denser and stronger than bone, the endoprosthesis was assigned a rigid body material to simplify the FEA.

Final implantation was modeled with a displacement ramp function simulating a quasi-static press-in condition over 1 s. The bone was fixed in all directions proximal to the surgical neck of the humerus and tibial tuberosity. The endoprosthesis began 5 mm outside the distal osteotomy (no contact with the bone) and was moved into place so contact began after 2 mm displacement and terminated when the distal end of the endoprosthesis was flushed to the distal osteotomy. The endoprosthesis was fixed in all degrees of freedom except along the long axis of the bone, which was aligned with the implantation axis of the endoprosthesis, and the endosteal surface of the medullary canal was prepared (**Figure 2**). A sliding elastic contact was assigned

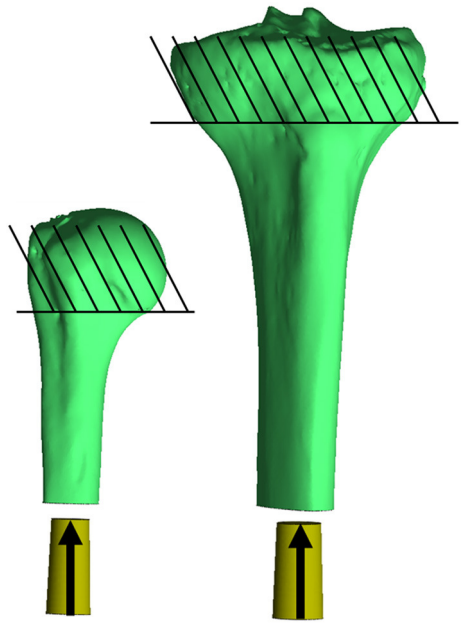

FIGURE 2 | Impaction of the endoprosthesis along the long axis of the bone through the centroid of the medullary canal, parallel to the inertial axis of the medullary canal. The endoprosthesis was placed at the distal end 5 mm proud the distal osteotomy, so no contact occurred with the bone (30 mm to the bone from where it is currently pictured). Contact began with the endoprosthesis 3 mm proud the distal osteotomy, and subsequent impaction created a maximum of ∼-0.1 mm mismatch in diameter of the prepared bone endosteal surface to the endoprosthesis. Bones were fixed proximal to the humeral head and tibial tuberosity. *Relative sizes of the bones are approximately to scale.

between the two objects with a coefficient of friction of 1.3 determined by testing the PODS porous coating (P^2 Porous Coating, DJO Surgical) on cancellous bone foam (23).

Cadaveric Testing

Physical impaction tests were completed on the three cadaver tibias to quantify the force of experimental impaction. Each bone was prepared according to the device manufacturer (8). Bone preparation stopped when the endoprosthesis could be placed in the medullary canal 3 mm proud the distal osteotomy. A part comparison analysis (conducted in 3-Matic) of CT reconstructions was performed to verify the accuracy of virtual implantation used for FEA models compared to prepared cadaver bones. The RMS error between the two surfaces was recorded.

Finally, constructs were loaded onto a material test machine (Model 858 Mini Bionix II; MTS Systems, Eden Prairie, MN, United States) with a 25-kN load cell (#622.2OH-05; MTS Systems, Eden Prairie, MN, United States) so that loading was along the long axis of the bone and endoprosthesis. By displacement control, the machine pressed the endoprosthesis in place at a rate of 5 mm/min, terminating at a set displacement measured with calipers between the distal osteotomy and endoprosthesis end loading collar. The speed was selected based on an initial evaluation of sampling rate on the force vs. time curve to ensure that accurate force was captured without extensive interpolation among points. Force and displacement data were acquired at 1 kHz.

Data Analysis

The final maximum rigid force of the endoprosthesis in response to the bone, corresponding to the impaction force, was recorded over the impaction period for FEA models. Additionally, circumferential stress, corresponding to the circumferential stress around the bone, was recorded. Sizes of all the endoprosthesis models were compared to the average diameter of the medullary canal at the distal osteotomy, and rounded to the nearest whole number for consistent comparison independent of medullary canal diameter and nominal endoprosthesis size.

The amount of bone-endoprosthesis contact at final implantation was also recorded as a percentage of the possible contact surface area of the endoprosthesis (20).

RESULTS

Finite Element Models

A total of 21 humerus models were constructed from the three bones with indicated endoprosthesis sizes of 10, 10, and 11 mm. These included endoprostheses barely contacting the endoprosthesis around the distal osteotomy and increasing until just before there was penetration through the periosteal cortex at the distal osteotomy. A total of 25 tibia models were created from the three bones with indicated sizes of 11, 14, and 15 mm and the same size range criteria. All cases were normalized to the average diameter for that particular bone (size 0), resulting in a size comparison from a minimum of −3 to maximum of +6 (**Figure 3**). Each step corresponds to a radial increase of 1 mm in the diameter of the endoprosthesis. The humerus and tibia models had an average of 349,451 and 542,684 elements, respectively.

Both the humerus and tibia models followed similar trends in impaction force and circumferential stress (**Figure 3**). Impaction force had a sharp increase to a maximum at a normalized endoprosthesis size of +0-1 for the humerus and +3 for the tibia. The force then decreased slightly as the endoprosthesis size continued to increase. The circumferential stress increased, plateaued, or decreased slightly, and then increased again for all the models (**Figure 3**). The plateau occurred at +0–2 for the humerus and +2–3 for the tibia before increasing again as endoprosthesis size increased (**Figure 3**).

Qualitative observations of the stress field revealed that the maximum stress was concentrated in thin-walled regions. This was more dispersed for the humerus (**Figure 4**) but concentrated in the medial and lateral regions for the tibia (**Figure 5**). Subject-specific morphologic features created smaller stress concentrations around the medullary canal, especially for smaller-size endoprostheses where these features were not removed by reaming (**Figures 4**, **5**). This was more pronounced in the tibia where the medullary canal was more elliptical than in the humerus, meaning a larger endoprosthesis was necessary before making contact around the circumference and removing more model-specific morphologic features.

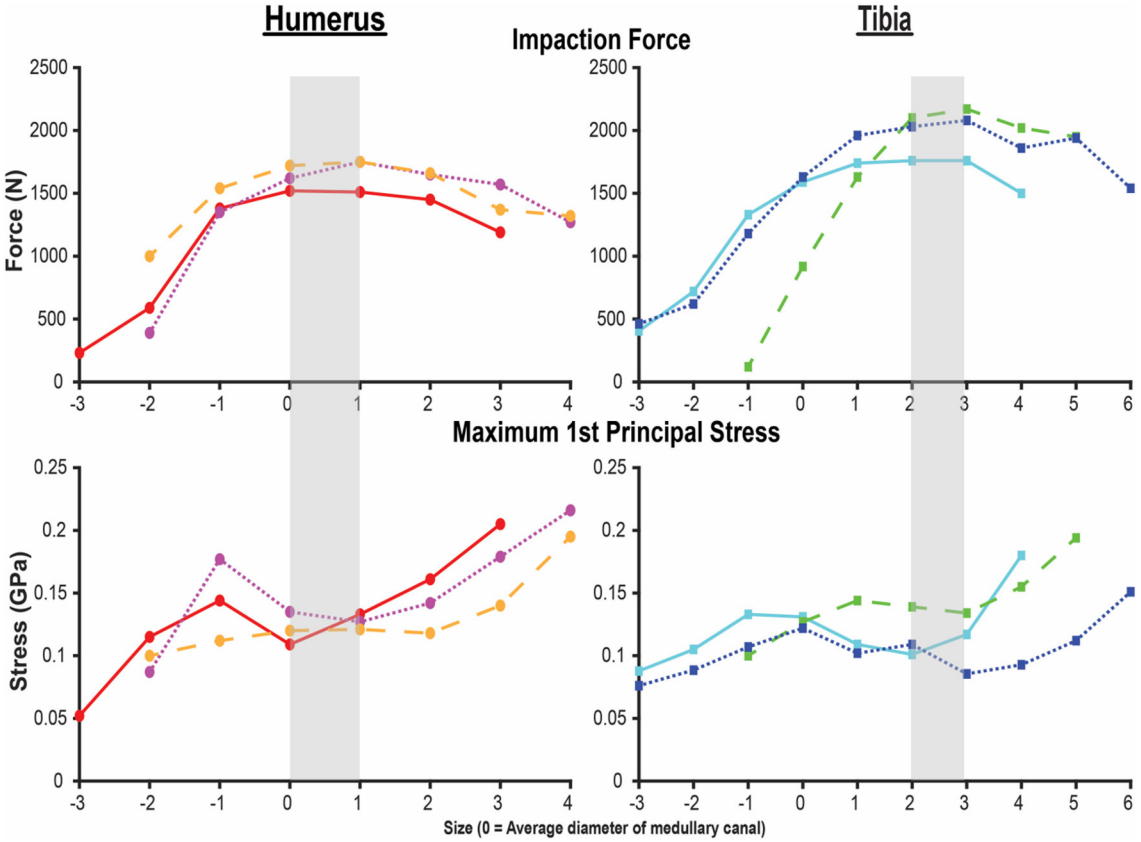

FIGURE 3 | Final impaction force (top) and maximum first principal stress (bottom) for the humerus (left) and tibia (right) finite element analysis (FEA) models. X-axis is the endoprosthesis size normalized to the average diameter of the medullary canal. Each line represents one of the three humeri and tibiae modeled by FEAs. Shading indicates sizing that maximizes impaction force while minimizing stress. Small peaks and valleys in the overall pattern occurred in areas of a subject-specific feature of the medullary canal.

Resultant contact area at final implantation revealed that humeral implantations achieved 58.4 ± 11.2% contact at the indicated size implant, while the tibia achieved 40.2 ± 26.1% bone-endoprosthesis contact (**Table 1**). The humerus achieved more than 13% bone-endoprosthesis contact for sizes −1–2 compared to the tibia.

Cadaveric Testing

A part comparison analysis revealed an average RMS error between surfaces (range) of 0.24 mm (0.15–0.33 mm). Testing revealed that the FE models overestimated impaction force by 334 ± 124 N (mean ± STD) (**Table 2**).

DISCUSSION

Our primary objective was to determine the mechanism that causes a large discrepancy in mechanical failure data of the same endoprosthesis for the humerus and tibia. We sought to evaluate the influence of endoprosthesis size on impaction force and circumferential stress in both anatomic locations. We hypothesized that the humerus and tibia would require a different sizing criterion in order to use the same PODS design to maximize impaction force and minimize circumferential stress. This hypothesis was confirmed, as the optimal sizing for the humerus corresponded to implantation instructions, whereas the optimal sizing for the tibia was three sizes larger than the instructions indicated. These results are specific to the press-fit, tapered porous-coated region with a circular cross-section of the PODS system and may not necessarily translate to other fixation approaches currently in use for percutaneous OI attachment systems, such as threaded screws (24). The results also varied with longer systems that apply a press fit to a bigger region of the bone (25).

During mechanical tests of PODS devices on the humerus and tibia of humans, fractures were observed along the long axis of the bone, primarily in thin-walled regions. Similar fracture patterns in the femur during impaction of total hip replacements (11–13) suggests failure due to circumferential stress arising from the dilation of the bone from endoprosthesis interference. The circumferential stress was analyzed and showed an intermediate plateau region with 12 ± 11% and 12 ± 5% stress variance in the humerus and tibia (**Figure 3**), respectively. This plateau occurred around sizes +0–2 or −1 to 1 in the humerus and +1–3 or +1–4 in the tibia. Failure stresses in radial dilation of cortical bone

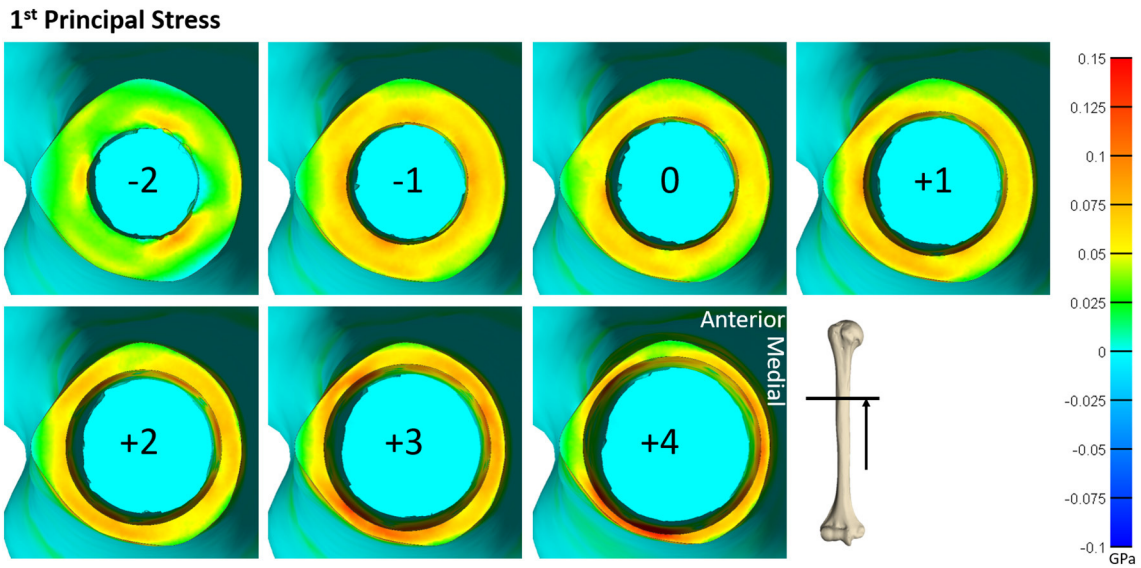

FIGURE 4 | Heat map of the first principal stress (circumferential stress) for each size of implanted endoprosthesis for one of the humerus models. For small endoprosthesis sizes, maximum stress occurred around subject-specific morphologic features. Once the endoprosthesis made contact with more areas of the bone (around size 0), a more uniform stress distribution was observed. Peak stress also began propagating through the thickness of the bone at the distal end around size +3.

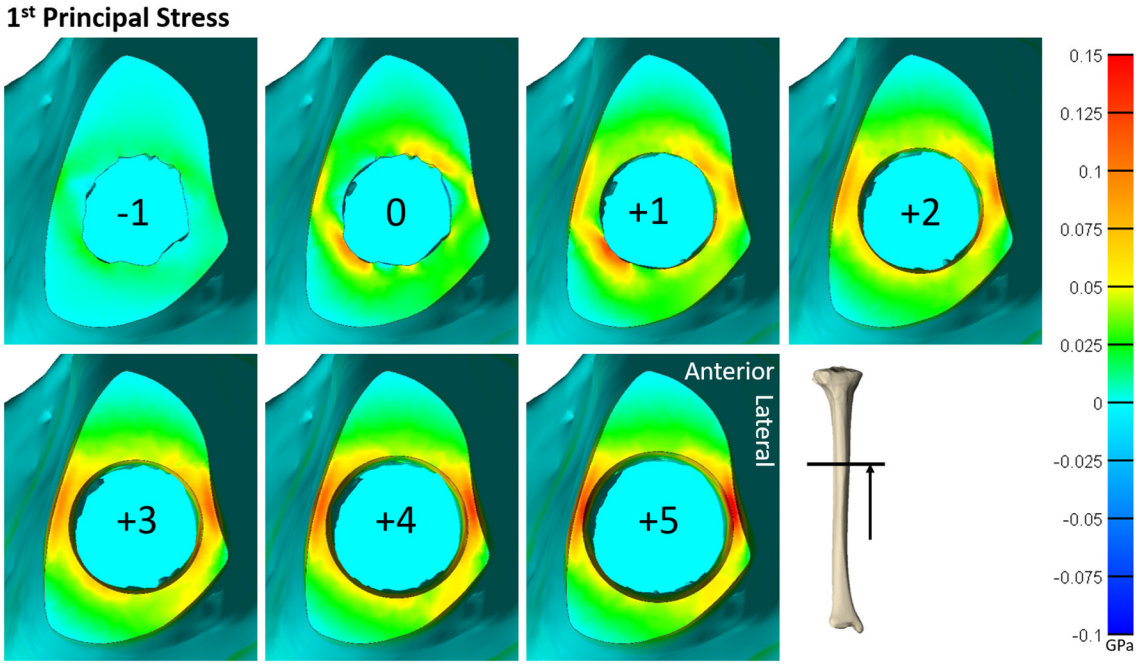

FIGURE 5 | Heat map of the first principal stress (circumferential stress) for each size of implanted endoprosthesis for one of the tibia models. For small endoprosthesis sizes, maximum stress occurred around subject-specific morphologic features. Once the endoprosthesis made contact with more areas of the bone (around size 3), peak stress was uniformly distributed around the medial and lateral regions, and began to propagate through the thickness of the bone at the distal end.

have not been well characterized. However, the most comparable study on ultimate stress performed compressive failure testing on bone plugs taken from the radial and circumferential axes of the femur, and found an ultimate stress of 0.063 and 0.065GPa, respectively (26). These results do not provide a direct measure of stress from radial dilation of the bone, since both sides are fixed, but they provide a fracture risk threshold in the correct loading direction of observed fractures. Other tests that have

TABLE 1 | Bone-endoprosthesis contact (%).

	Bone-endoprosthesis contact (%)					
Normalized size	**Tibia 1**	**Tibia 2**	**Tibia 3**	**Humerus 1**	**Humerus 2**	**Humerus 3**
−4		5.4				
−3	7.9	13.4		6.8		
−2	20.7	22.9		21.8	17.6	3.7
−1	37.5	34.7	1.2	41.4	44.9	27.0
0	58.9	51.3	10.4	64.7	65.1	45.6
1	76.7	70.7	27.0	83.6	83.9	65.1
2	93.7	89.5	49.4	100.0	99.7	84.2
3	100.0	100.0	70.8	100.0	100.0	100.0
4	100.0	100.0	91.2		100.0	100.0
5		100.0	100.0			
6		100.0				

Amount of bone in contact with the endoprosthesis as a percentage of the total possible contact surface of the endoprosthesis.

TABLE 2 | Results of impaction force from FEA and cadaveric tests on the same tibia bone.

Tibia specimen	**Average Young's Modulus (GPa)**	**Final impaction force (N)**	
		FEA	**Cadaver**
1	11.02	2,101	1,741
2	10.04	1,958	1,758
3	9.45	1,326	884

Average Young's Modulus was calculated from calibrated CT scans of cadaver bones used in mechanical testing and informing material properties of elements in the bone mesh for FEA.

characterized transverse properties of cortical bone report an ultimate stress of 0.131 GPa (27). Almost every case examined by FEA had a maximum circumferential stress within this range (**Figure 3**). However, below a normalized size of +3, this was a localized point around a specific morphologic feature of the medullary canal and did not propagate through to the outer cortex (**Figures 4**, **5**). Stress propagation to the outer cortex would increase the risk of periprosthetic fracture instead of localized fracture to a small intramedullary feature. Targeting sizes before heightened stress propagation and in the plateau of lower stress is ideal to minimize stress while maintaining high impaction force.

Higher impaction force correlates to tighter fit and increased initial stability (10) of the bone-endoprosthesis interface as long as it does not create too much stress and increase risk for fracture. In the humerus, maximum impaction force occurred in sizes +0–1, meaning the average diameter of the medullary canal at the distal osteotomy is a good indicator of endoprosthesis size that maximizes impaction force. In the tibia, this occurred in size +3. There is a large difference between these two bones in decrease in impaction force after the maximum. For the humerus, the difference in impaction force from size +1 to +2 was only 83 ± 21 N (4.9 ± 0.9%). In the tibia, the difference in impaction force from size +3 to +4 was 210 ± 56 N (10.8 ± 3.9%). This indicates that there is little room for error when trying to achieve maximum impaction force while not fracturing the bone, especially in the tibia at this examined amputation level.

In the humerus, the stress plateau coincided with peak impaction force (**Figure 3**). In the tibia, peak impaction force was at the high end of the plateau just before the sharp increase in stress (**Figure 3**). As a result, there is more room for error to achieve the maximum impaction force with a smaller stress for the humerus compared to the tibia, providing a possible explanation as to why the mechanical testing results (**Supplementary Materials**) (8) were so different between the bones, beyond the fact that optimal sizing is different. In mechanical testing of the tibia, endoprostheses were three sizes smaller than the optimal size the FEA predicted, decreasing impaction force and associated stability.

The humerus experienced a more uniform stress distribution around the circumference of the endoprosthesis because of the uniformity of cortical thickness (**Figure 6**). The tibia had very thin-walled medial and lateral sides and concentrated stress to specific regions with larger endoprosthesis sizes (**Figure 6**). This localized high stress in the tibia highlights a pattern that would benefit from a different design approach that preserves the medial and lateral cortices to maintain cortical thickness during impaction, like an elliptical cross-section (28). To test this concept, we modeled a tapered, elliptical endoprosthesis with a major diameter 2 mm larger than the diameter of the circular cross-section geometry, aligned to the anteroposterior axis, and held all other parameters constant. We then implanted it into one of the modeled tibias and included the same array of endoprosthesis sizes, with revised preparation so that interference would remain the same for the new shape (**Figure 7**). This pilot test maintained circumferential stress but showed a 489 ± 62 N increase in impaction force for all endoprosthesis sizes (**Figure 7**). This increased initial stability is due, in part, to the achievement of more contact with the bone and preservation of thin cortex regions creating more resistance to deformation and more uniform distribution of the stress around the endosteal surface. Further research should determine how aggressive this

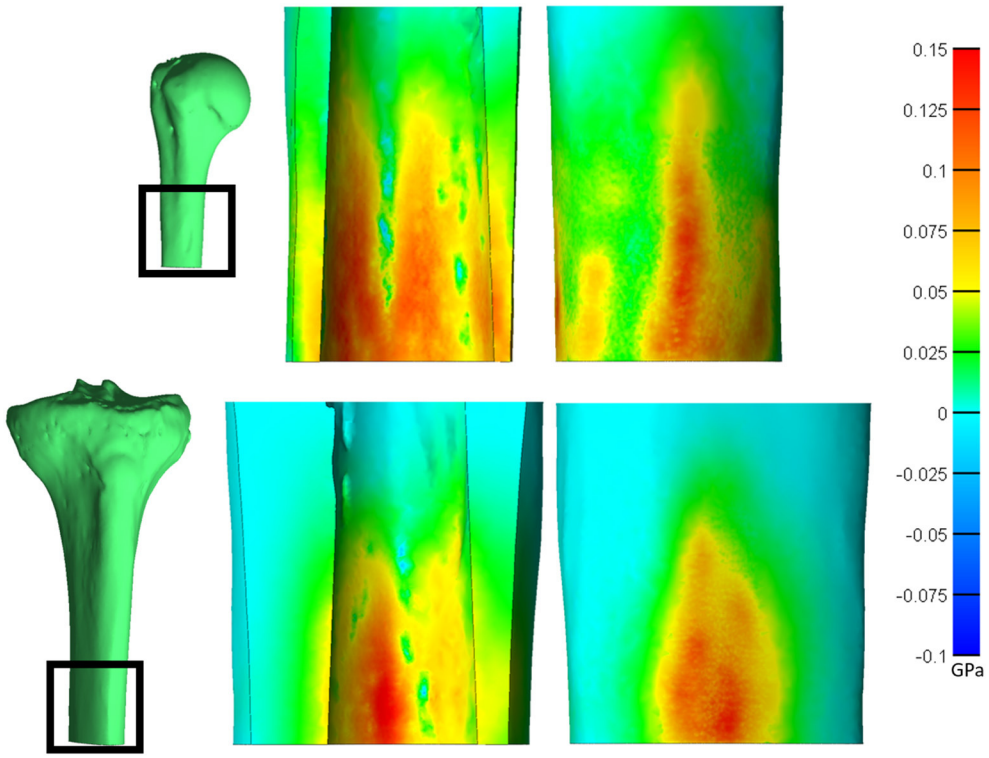

FIGURE 6 | Saggital cross-section of the humerus (top) and tibia (bottom) showing the endosteal (middle) and periosteal (right) surfaces. Heat map represents the first principal stress. Both bones have a normalized size of +3. *Relative sizes of the bones are approximately to scale.

ellipse should be to accommodate the morphology of the tibia. Based on the current results, we hypothesize that increasing the major diameter of the endoprosthesis would increase impaction force without significant change in stress until the cortex in the anterior and posterior regions are thinned similar to the medial and lateral regions. Further analysis is required to confirm this hypothesis. Clinically, the surgical instrumentation necessary to achieve this elliptical shape would be more difficult to design. We propose using the same reamer geometry followed by a broach with increasing major diameter to achieve this cross-sectional shape.

Other cross-sectional geometries may also perform better than the circular cross-section, and the highly variable morphology of the tibia along the long axis may benefit from several amputation-level specific design approaches. However, this increases implantation cost with multiple tooling sets. This problem needs to be addressed and balanced to optimize mechanical stability while allowing for implantation in a wide population of individuals with transtibial amputation.

Validation of the model is necessary to confirm the observed stress field and make more specific claims about how this endoprosthesis design performs. This validation should include strain gauge measurements or optical tracking of strain to validate beyond impaction force to inform the model. The cadaveric testing showed that the current model parameters were of good approximation but overestimated the impaction force by 333.8 $\pm$ 123.6 N (**Table 2**). This could have been influenced by the use of four-node tetrahedral elements that are a more rigid element type and the fact that the coefficient of friction used is determined by testing cancellous bone foam due to the unavailability of a value determined on cortical bone. Besides tuning model parameters, there were factors in the cadaveric testing that could cause the disagreement. The models also did not simulate the porous coating on the endoprosthesis that files away bone when impacted. Also, the actual endoprostheses have a diametric variation of the porous coating, meaning the size modeled in the FEA might not exactly match that used in cadaver tests. Additionally, experimentally measuring 3 mm proud the distal osteotomy with calipers adds error in the preparation, because interference between the endoprosthesis and bone is not perfect. This could increase forces if the endoprosthesis was more than 3 mm proud and decrease them if less.

This study is limited in that the models were constructed based on a small sample size of non-amputee bones. Heterotopic ossification, osteoporosis, cortical thinning due to disuse atrophy, and other changes in bone morphology are common for lower extremity amputees (29–31), and would decrease the impaction force and stress in patient populations with lower bone quality. The full range of endoprostheses that did not penetrate the periosteal surface was examined to try to capture the case of very thin cortex possible for amputees with disuse atrophy of the bone. A small, representative sample size was selected to begin to

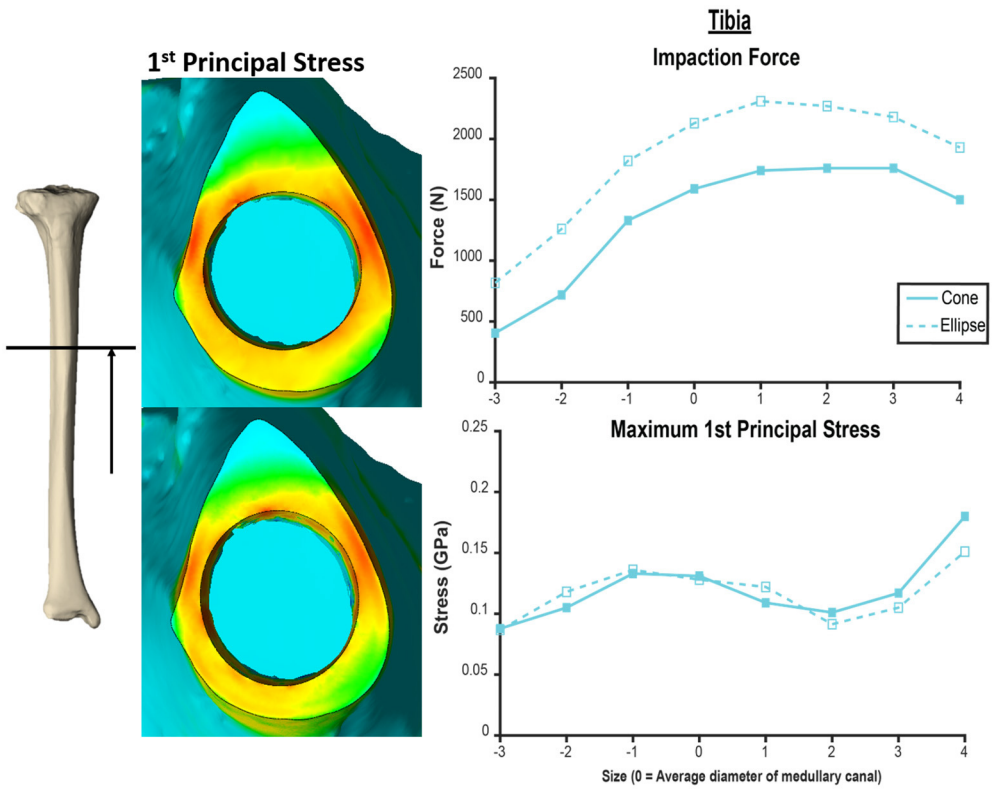

FIGURE 7 | Comparison of tapered endoprosthesis with circular cross-section (left top, right solid) and endoprosthesis with tapered elliptical cross-section (left bottom, right dashed). The elliptical endoprosthesis has a 1-mm larger radius of major and minor diameters. The minor diameter of the ellipse matches the diameter of the circular endoprosthesis. This comparison was conducted on the same tibia bone model. Heat map scale of first principal stress (left) is uniform across both models.

elucidate the mechanisms causing observed mechanical failure. We did not employ statistical shape modeling for bone geometry, because we also performed mechanical testing on the same cadaver bone and used the CT scan of the cadaver to determine element-specific mechanical properties. This correlation between mechanical testing results and FEA models would not be possible with a statistical shape model. Once a revised endoprosthesis has been designed that improves on previous initial stability results, FEAs using statistical shape models would be beneficial to assess the new design on a wider population.

Furthermore, only one amputation level for both the humerus and tibia was modeled. Investigation of more amputation levels would determine if these findings indicate a new design in the tibia is applicable to the entire length of the bone. At amputation levels more distal to 40%, a circular cross-section may be suitable, since there is a more uniform circular medullary canal and cortex (9), but long amputations may not present as good candidates for percutaneous OI attachment because of prosthetic component height (20). Additional studies are necessary to refine FEA models and investigate device designs before implementing a percutaneous OI endoprosthesis into the population of transtibial amputees.

This study modeled the impaction of the PODS porous-coated OI region in the humerus and tibia by FEA. Forces and circumferential stresses were recorded for impaction with an array of endoprosthesis sizes, revealing that current implantation protocols are optimized for transhumeral implantation but not for tibial implantation. The tibia requires an endoprosthesis with a diameter larger than previously predicted for the same PODS OI region in order to achieve maximum impaction force, but this quickly causes an increase in periprosthetic stress. In order to achieve safe implantation of a transtibial endoprosthesis, we recommend further investigation on an endoprosthesis with elliptical cross-section based on the preliminary investigation of this device and failure to achieve acceptable results with the current PODS system.

DATA AVAILABILITY STATEMENT

The raw data supporting the conclusions of this article will be made available by the authors, without undue reservation.

AUTHOR CONTRIBUTIONS

CT, HH, and KB contributed to the design and execution of this study. CT carried out data analysis and initial

manuscript authorship. HH and KB contributed to the manuscript with major edits and revisions. All authors contributed to the article and approved the submitted version.

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SUPPLEMENTARY MATERIAL

The Supplementary Material for this article can be found online at: https://www.frontiersin.org/articles/10.3389/fresc.2021.744674/full#supplementary-material

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frontiers in Rehabilitation Sciences

CASE REPORT
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Case Report: Bionic Reconstruction in an Adult With Obstetric Brachial Plexus Injury

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Introduction: Many adults who had a severe Narakas IV obstetric brachial plexus injury (OBPI) suffer from extensive impairments in daily living due to limited hand-arm function. The dramatic loss of axonal support at this very early age of development often render the entire extremity a biologic wasteland and reconstructive methods and therapies often fail to recover any functional hand use. In this scenario bionic reconstruction, including an elective amputation and a subsequent prosthetic fitting, may enable functional improvement in adults suffering from the consequences of such severe brachial plexus injuries. We here describe our experience in treating such patients and lay out the surgical rational and rehabilitation protocol exemplified in one patient.

Case Presentation/Methods: A 27-year-old adult with a unilateral OBPI contacted our center. He presented with globally diminished function of the affected upper extremity with minimal hand activity, resulting in an inability to perform various tasks of daily living. No biological reconstructive efforts were available to restore meaningful hand function. An interdisciplinary evaluation, including a psychosocial assessment, was used to assess eligibility for bionic reconstruction. Before the amputation and after the prosthetic fitting functional assessments and self-reported questionnaires were performed.

Results: One month after the amputation and de-rotation osteotomy of the humerus the patient was fitted with a myoelectric prosthesis. At the 1.5 year-follow-up assessment, the patient presented with a distinct improvement of function: the ARAT improved from 12 to 20 points, SHAP score improved from 8 to 29, and the DASH value improved from 50 to 11.7. The average wearing times of the prosthesis were 5 to 6 h per day (on 4–5 days a week).

Discussion: The options for adults suffering from the consequences of severe OBPIs to improve function are limited. In selected patients in whom the neurological deficit is so severe that biologic hand function is unsatisfactory, an elective amputation and subsequent restoration of the hand with mechatronic means may be an option. The follow-up results indicate that this concept can indeed lead to solid hand function and independence in daily activities after amputation, subsequent prosthetic fitting, and rehabilitation.

Keywords: obstetric brachial plexus injury (OBPI), bionic reconstruction, upper limb amputation, prosthesis, case report, functional outcome

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INTRODUCTION

Obstetric brachial plexus injuries (OBPI) refer to injuries of the brachial plexus that occur during delivery (1). The incidence of OBPI is documented in Norway with 0.3%, with a relatively high recovery rate, nonetheless one in every 2,000 babies has to live with a permanent injury of the plexus (2). Guidelines for patients with OBPI recommend early referral to multidisciplinary centers (at 1 month of age) (3). If no recovery occurs, early surgery is indicated at 3–9 months after birth, depending on the extent and severity of the injury (4).

If these early interventions do to not lead to sufficient outcomes, only a few surgical interventions are available after adolescence. They include the modified Quad surgery (5), tendon transfers for restoration of external shoulder rotation, and humeral rotational osteotomy in combination with lengthening (6). These limited options are reflected by the impaired hand function described by many adults after severe OBPI. Common clinical findings include problems in performing daily activities due to a lack of useful hand function, a high prevalence of pain as well as reduced sensation, arthritis, and an overall reduced quality of life (1, 7, 8). Despite their perceived disability, this patient group rarely receives rehabilitation measures (1) which might be related to the limited options available.

Recently, the method of "bionic reconstruction" has expanded options for patients with a very limited upper limb function. The procedure includes elective amputation of the hand, de-rotation osteotomy of the humerus for better positioning of the forearm, followed by prosthetic fitting. The feasibility of bionic reconstructions in patients suffering from brachial plexus injuries in adulthood is well-documented (9–11). However, studies investigating this treatment after OBPI are not found in literature. The aim of this report is to present a further indication for this procedure in patients who suffer the consequence of severe birth related plexus lesions. We report the case of a young patient with a unilateral OBPI who underwent bionic reconstruction, including long-term functional outcomes.

CASE DESCRIPTION

In July 2019, a 27-year-old adult with history of a right-sided Narakas IV OBPI contacted our center with the wish for bionic reconstruction. The patient described himself as male. In the first year after his birth, reconstruction of the brachial plexus was performed, including direct replantation of the lower roots C8 and T1 to the spine. An improvement of function was documented in his medical report. During childhood and adolescence, the patient did not receive any therapeutical interventions regarding his OBPI. He was unsatisfied with his situation when approaching our center and described his arm as an "annoying appendix being in his way."

METHODS

After the patient presented at our center, different possibilities were discussed, and it was decided that further biological reconstructive efforts would not lead to favorable outcomes and that bionic reconstruction should be evaluated. Therefore, the previously established guidelines for the procedure including a psychological assessment were followed (12, 13). Inclusion and exclusion criteria for bionic reconstruction have been described previously (11, 13). Furthermore, the patient received 6 days of intensive rehabilitation and home training by a physical and occupational therapist (AS and AB) with details outlined below and depicted in **Figure 1**. The patient gave written informed consent to take part in this study and standardized guidelines for reporting the case report (CARE checklist) were used (refer to **Supplementary Material 1**).

Clinical Examination

The right arm presented hypoplastic with an internal rotation deformity at shoulder level and flexion contracture in the elbow (see **Figure 2A**). The fingers and thumb were fixed in a flexed position, but minimal flexion of the thumb was possible. The patient was able to clamp small objects (for instance a wooden cube $2 \times 2 \times 2$ cm) between his thumb and fingers, however, had issues releasing them. There was minimal active movement of the wrist in extension and flexion. Active shoulder abduction was 80° and flexion was 110°. Active elevation of the arm with evasion movement was possible to 150°. The elbow showed a passive extension deficit of 75° and active flexion of up to 100°. The patient presented without any useful sensation in the hand and forearm.

Surface Electromyography Biofeedback Training and Training With a Table-Top Prosthesis

In a first step, surface electromyography (sEMG) signals on the forearm were identified [following established protocols (12, 14)] (by AS and AB). This was done by using an sEMG biofeedback system, where the muscle activation could be observed on a screen. Various electrode positions on the forearm and movement commands (like closing the fingers, flexing the wrist, opening the hand, extending single fingers, etc.) were tested. The aim was to find two different sEMG signals, one signal for opening the prosthesis and another for closing the hand. After the identification of the most appropriate electrode positions and movement cues (for the patient the best cues were flexing the fingers and extension of the thumb), these were trained separately. The patient was asked to activate one signal while the other remained relaxed and vice versa with a rest period in between. As soon as the activation of the signals could be reliably performed, the movements were practiced with a table-top prosthesis (opening and closing of the hand). This allowed the patient to receive direct feedback regarding movement intention and subsequent prosthetic motion.

Fitting and Training With Hybrid Prosthesis

A hybrid prosthesis that could be attached on the paretic arm with the pre-defined electrode positions was initially fitted (see **Figure 2B**). Intensive training with the device (12, 14) started with opening and closing of the hand in various speeds and different positions (standing, sitting, different arm positions). In a second step, grasping and manipulation of objects was trained

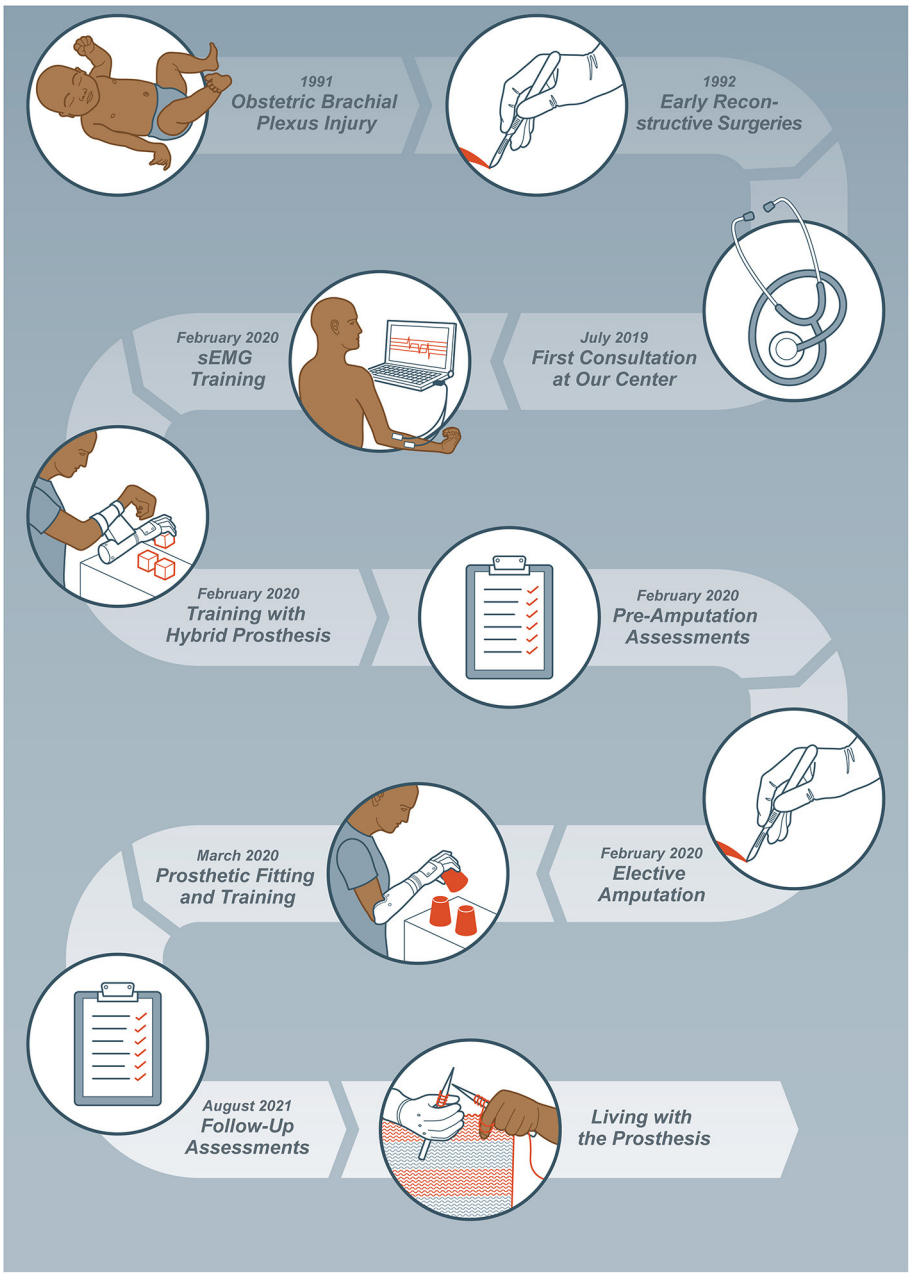

FIGURE 1 | Timeline of the patient undergoing bionic reconstruction.

in therapy and at home. Finally, simple tasks of daily living could be trained with the hybrid prosthesis, as a proof-of-concept. This also allowed the patient to experience limitations of current prosthetic devices (such as the lack of sensory feedback) before final decision making. Additional effects of the training were strengthening of the biceps and shoulder muscles.

Psychosocial Assessment

In a semi-structured interview the psychologist (AP) assessed the overall psychosocial status, the patient's motivation for the amputation, and the expectations of the outcome following the guidelines outlined by Hruby et al. (13). One of the major points thereby always addressed is the fact that an amputation is an irreversible procedure and that a prosthesis is only a tool which cannot be compared with an intact biological hand. As the patient was assessed as psychologically stable, not meeting any exclusion criteria and being aware of the consequences of the procedure, clearance for the planned amputation was given.

Functional Assessment

The current status and function of the arm and hand were assessed with standardized assessments including the

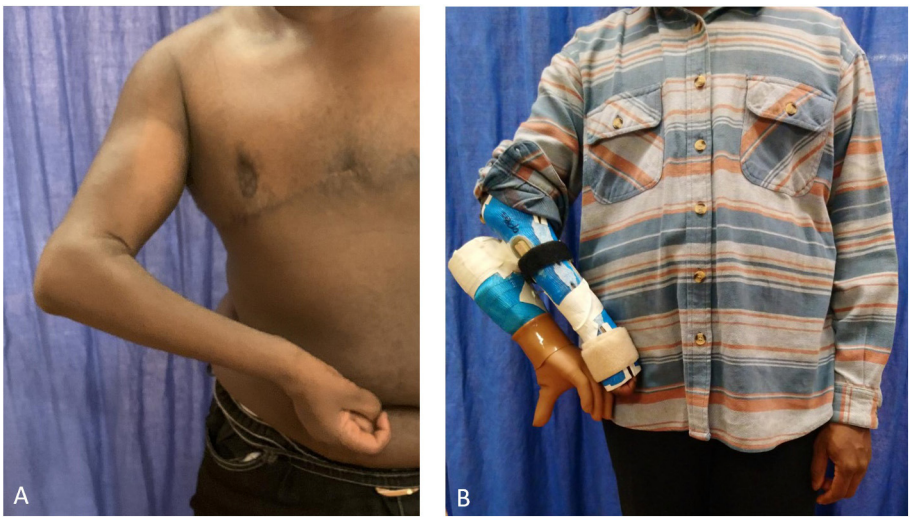

FIGURE 2 | Plexus arm **(A)** and hybrid prosthesis that is mounted on the paretic arm **(B)**.

"Action Research Arm Test" (ARAT) and "Southampton Hand Assessment Procedure" (SHAP) by a physical and occupational therapist (by AS and AB). The ARAT assesses the function of the impaired upper limb in 19 tasks using grasping and manipulation of various objects, as well as gross motor movements. The highest score, indicating no impairment, is 57 and the lowest score, indicating no function, is 0 (15). The SHAP test is designed to assess prosthetic function, also including grasping and manipulation of objects and tasks of daily living (such as opening a jar, undoing buttons, etc.). The time for each task is measured and determines the overall functional score, with 100 indicating normal function and 0 indicating no function (16). Both tests were conducted in a standing position. First, the native function of the affected hand was tested, afterwards the same tests were conducted with the hybrid prosthesis attached. As these functional tests using the hybrid prosthesis indicated acceptable prosthetic control, amputation was considered suitable from a functional perspective as well. Additionally, the "Disabilities of the Arm, Shoulder and Hand" (DASH) questionnaire, which assesses the limitations in everyday life due to an injury of the upper extremity, was completed by the patient. Here, 0 corresponds to no disability and 100 shows a complete dependence in daily life (17, 18). The current pain level was documented as well by using the visual analog scale (VAS) (100 mm line). 1.5 years after the final prosthesis fitting these assessments were performed again and a semi-structured interview was conducted. Additionally, some open questions and rating questions on an 11-level numeric rating scale (NRS) (0 means disagree/never and 10 agree/always) regarding the use of the prosthesis and the prosthesis embodiment were asked (11).

Surgery

After the approval from the multidisciplinary team the surgery took place in the same month. The procedure included a de-rotation osteotomy of the humerus, a shortening of the olecranon to release the extension deficit in the elbow and the transradial amputation (performed by OCA).

Prosthetic Fitting and Prosthesis Training

The rehabilitation process was started by a rehabilitation physician in the home country of the patient in March 2020. The rehabilitation team consisted of occupational/physical therapists, prosthetist and physician. After the surgical wounds had healed the patient received a prosthetic fitting with a MyoHand VariPlus Speed (Ottobock, Duderstadt, Germany) (see **Figure 3**). He attended a weekly prosthetic training (30 min per session) by an occupational/physical therapist for ∼1 year and trainings ongoing. The therapy consisted of simple movements of the prosthesis (opening/closing) in different speeds and positions, and further training of grasping and manipulation of different abstract objects. In a last step activities of daily living were trained with the prosthesis (including knitting). Also exercises for strengthening, endurance and symmetry of the body were discussed. The patient stayed in contact with our team *via* email and video calls.

RESULTS

As shown in **Table 1**, the ARAT improved from 12 pre-operatively to 20 at 1.5 years after the surgery, the SHAP test from 8 to 29 and the DASH showed an improvement from 50 to 11.7 (raw data can be found in the **Supplementary Material 2**, videos from one ARAT task can be found in the **Supplementary Material 3**). The testing with the hybrid prosthesis had already indicated an improvement of hand function compared to the biological arm with an ARAT score of 17 and a SHAP score of 19. The patient described no pain before and after the bionic reconstruction. The extension deficit of the elbow improved through treatment, with a final range of motion of 0°-55°-100°.

FIGURE 3 | Patient with the final prosthetic fitting 1.5 years after the amputation.

TABLE 1 | Results of the assessments with the plexus hand, the hybrid prosthesis and the final prosthesis (1.5 years follow-up).

	Plexus hand	Hybrid prosthesis	1.5 years follow-up with prosthesis
DASH	50	-	11.7
ARAT	12	17	20
SHAP	8	19	29
Pain (VAS)	0	-	0
ROM elbow flexion	0°-75°-100°	-	0°-55°-100°

A higher score for the ARAT and SHAP test shows a better function whereas a lower score in the DASH questionnaire shows less disabilities. 0 means no pain on the visual analog scale (VAS). Because of the flexion contracture the range of motion (ROM) of the elbow is presented as no extension possible (0°)—minimal flexed position to maximum flexed position.

The average wearing time of the prosthesis was 5–6 h per day (on ∼4–5 days per week) and the patient reported a particular preference of wearing the prosthesis when leaving his house. The patient further reported to sometimes wear his prothesis switched off, activating it only when needed. He liked wearing the prosthesis a lot (NRS "9/10"). "I did bimanual tasks with my intact arm/hand together with my prosthesis" was rated by the patient with "9/10" (NRS). He had the feeling that the prosthesis was a part of the body (NRS "10/10") and that his prosthesis looked like a real part of the body (NRS "9/10"). Also the statement "I felt the prosthesis only as a tool, and not as a part of my body" was rated with "9/10" on the NRS. "I felt that I had full control over the prosthesis" was rated with "9/10" on the NRS (also see **Supplementary Material 2**). The satisfaction of the current function was rated with "5/10" on the NRS. In the personal interview at follow-up, he described that he sometimes struggled to control the prosthesis and that he wished for a different prosthesis model (a multi-articulating hand with different grasping types). Nevertheless, with the prosthesis he was able to do things that were not possible before, such as holding and fixing objects, carrying a bag and even knitting. He perceived his quality of life much higher than before and told us he would undergo the procedure again. If he had the choice, he might even opt for bionic reconstruction sooner. No adverse or unanticipated events occurred during the process.

DISCUSSION

The clinical prognosis after OBPI depends particularly on the severity and extent of the injury (as classified by Narakas), early surgical interventions if needed, and subsequent rehabilitation (3). While the majority of patients develop good upper extremity function (19), in some cases the motoneuron loss is of such extent that the entire neuromuscular system will undergo fatty-fibrous degeneration leading to multiple joint contractures and deformities rendering the extremity with severe impairments and reduced quality of life (1, 7, 8). This was the situation of a 27-year-old adult who approached us for consultation in 2019. He reported a great disability in daily life due to a Narakas IV OBPI, with resulting socioeconomic limitations such as inability to complete nursing school. The efforts that had already been pursued to improve the situation did not lead to satisfactory results for the patient. Furthermore, surgical procedures to improve function in severe Narakas IV lesions of the plexus are limited and restoring meaningful hand function is challenging (4). A case report of three female adults undergoing a modified Quad surgery, which is a combination of muscle transpositions, resulted in an improvement of the total modified Mallet Score in two of them after the surgery (5). Another case report could demonstrate an improvement of shoulder function after an external rotation osteotomy and lengthening of the humerus documented with the modified Mallet Score (6). In both studies the impact of the intervention on hand and arm function in daily life activities was not explored. Overall, outcomes for hand function after secondary brachial plexus reconstruction are very limited from a functional perspective (20, 21). Tendon transfers were deemed not feasible due to a lack of local muscles for hand reanimation. A free gracilis transfer was discussed, however, omitted due to a lack of strong motor nerves for reinnervation, the contracted position of the hand as described above and, finally, the distinct wish of our patient against further reconstructive efforts.

For these reasons the possibility of elective amputation and subsequent prosthetic fitting was explored further with our

patient. While this procedure had not been described previously in adult patients with OBPI, its benefits are reported for other patient groups. For instance, a patient suffering from arthrogryposis multiplex congenita showed an improvement of function, daily activities, independence and quality of life after prosthetic reconstruction. The prosthesis enhanced his self-confidence in terms of his appearance, which promoted enjoyment of social interactions and activities (22). Similar outcomes are reported for patients who had undergone a bionic reconstruction after severe traumatic brachial plexus injuries (9, 10). In a first case series of three patients with brachial plexus injuries where the amputation was at a transradial level the mean ARAT score (±standard deviation) increased from 5.3 ± 4.7 to 30.7 ± 14, the mean SHAP from 9.3 ± 1.5 to 65.3 ± 19.4 and the mean DASH improved from 46.5 ± 18.7 to 11.7 ± 8.4 (10). Also for five patients with more severe brachial plexus injuries who underwent an amputation above the elbow with following prosthetic fitting the mean ARAT increased from 0.6 ± 1.3 to 17.3 ± 1.5, the mean SHAP from 4 ± 3.7 to 22 ± 9.2, the mean DASH decreased from 52.5 ± 9.4 to 31.2 ± 9.8 and the mean VAS from 8.5 ± 1 to 6.7 ± 2.1 (9). Moreover, the ability to act bimanually had a positive influence on the well-being of the patients and their social interaction with others (10). Altogether, the results of these studies are comparable with our case report, indicating that bionic reconstruction can improve function and independence in daily life, as well as have a positive effect on the quality of life in selected patients.

Comparable to these other indications, the described interventions resulted in an increase of hand and arm function and enabled our patient to perform tasks of daily living, which were not possible beforehand. In contrast to patients with a brachial plexus injury in adulthood, our patient never experienced his affected hand as functional. Considering this, it is remarkable that despite the life-long lack of hand function, the patient seemed to incorporate the prosthesis very well and indicated a high level of embodiment in the questionnaire. Interestingly, the statements "I had the feeling that the prosthesis was part of my body" and "I felt the prosthesis only as a tool, and not as a part of my body" were both rated very high on the NRS. The patient explained that in his view these two statements do not exclude each other in his perception and merely depend on the situation. When he is outside and interacting with others in social situations, the prosthesis is part of his body and gives him a feeling of bodily integrity. However, when he focuses on using the prosthesis during specific functional tasks, he perceives it as a tool. Furthermore, he stated that his prosthesis is definitely not the same as the biological hand, but still belongs to him. These observations reveal the complexity of body image concepts in this specific group of patients who sacrifice parts of their (non-functional) insensate human frame for a bionic replacement. These findings indicate the importance of investigating the topic of body image and embodiment with a combination of quantitative and qualitative research methods. Comparing the quantitative scores of the prosthetic embodiment with other patients who underwent bionic reconstruction after traumatic brachial plexus injury (11), our patient had higher ratings in all items *("I had the feeling that the prosthesis was part of my body.", "I felt the prosthesis only as a tool, and not as a part of my body.", "I did bimanual tasks with my intact arm/hand together with my prosthesis.", "I felt that I had full control over the prosthesis.", "I liked wearing the prosthesis.", "I felt that my prosthesis looked like a real part of the body.")*. These findings could be supported by the fact, that he now for the first time in life has a meaningful functional hand compared to the other cohort. In line with these ratings, using the prosthesis in gestures indicating a strong embodiment could be observed during the follow-up visit. They included touching the prosthesis with the unaffected hand, holding both "hands" and putting both "hands" in the trouser pocket.

The long follow-up period of 1.5 years after amputation is a strength of this case report. This period gives a good insight into the long-term outcomes of the final prosthesis use. In addition, the choice of assessments and questionnaires allows a holistic/exhaustive picture of the outcomes. The assessments were performed and scored by two experienced therapists and video recorded to increase reliability.

As this is a case report, we cannot generalize the results obtained from this one patient, however it does provide evidence that in severe cases of OBPI this concept will provide solid hand function with a high level of embodiment.

Patient selection as well as education and professional support through an experienced multidisciplinary team (including surgeons, occupational/physical therapists, psychologists and prosthetists) during the whole process are essential. A tailored psychosocial assessment and a structured rehabilitation program have proven very helpful in our experience. However, bionic reconstruction should only be performed, when biological restoration and rehabilitation measures have been exhausted and no other option is available. More research should explore the reconstructive and rehabilitative options for adults suffering OBPI, as this patient cohort is currently underrepresented in literature.

PATIENT PERSPECTIVE

The patient's quality of life has improved, as he is able to do things with the prosthesis he could not do before. Accordingly, he feels assured in his decision and would undergo bionic reconstruction again, maybe even at an earlier point.

DATA AVAILABILITY STATEMENT

The original contributions presented in the study are included in the article/**Supplementary Material**, further inquiries can be directed to the corresponding author.

ETHICS STATEMENT

The studies involving human participants were reviewed and approved by Ethics Committee, Medical University of Vienna.

The patients/participants provided their written informed consent to participate in this study. Written informed consent was obtained from the individual(s) for the publication of any potentially identifiable images or data included in this article.

AUTHOR CONTRIBUTIONS

CG, AS, OCA, and AB conceived and designed the study. AS, AB, AP, and CG performed the data acquisition and interpreted the data. AB, AS, CG, AP, OCA, and GL wrote and edited the manuscript. All authors gave final approval for publication.

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SUPPLEMENTARY MATERIAL

The Supplementary Material for this article can be found online at: https://www.frontiersin.org/articles/10.3389/fresc.2021.804376/full#supplementary-material

Supplementary Material 1 | CARE checklist.

Supplementary Material 2 | Raw data of DASH, ARAT and SHAP before the amputation, with the hybrid prosthesis and with the final fitting and raw data of the embodiment questions.

Supplementary Material 3 | Videos from ARAT Exercise "Block, 5 cm^3" with the plexus hand, the hybrid prosthesis and the final prosthesis (1.5 years follow-up).

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Free-Living User Perspectives on Musculoskeletal Pain and Patient-Reported Mobility With Passive and Powered Prosthetic Ankle-Foot Components: A Pragmatic, Exploratory Cross-Sectional Study

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Introduction: Studies with a powered prosthetic ankle-foot (PwrAF) found a reduction in sound knee loading compared to passive feet. Therefore, the aim of the present study was to determine whether anecdotal reports on reduced musculoskeletal pain and improved patient-reported mobility were isolated occurrences or reflect a common experience in PwrAF users.

Methods: Two hundred and fifty individuals with transtibial amputation (TTA) who had been fitted a PwrAF in the past were invited to an online survey on average sound knee, amputated side knee, and low-back pain assessed with numerical pain rating scales (NPRS), the PROMIS Pain Interference scale, and the PLUS-M for patient-reported mobility in the free-living environment. Subjects rated their current foot and recalled the ratings for their previous foot. Recalled scores were adjusted for recall bias by clinically meaningful amounts following published recommendations. Statistical comparisons were performed using Wilcoxon's signed rank test.

Results: Forty-six subjects, all male, with unilateral TTA provided data suitable for analysis. Eighteen individuals (39%) were current PwrAF users, whereas 28 subjects (61%) had reverted to a passive foot. After adjustment for recall bias, current PwrAF users reported significantly less sound knee pain than they recalled for use of a passive foot (−0.5 NPRS, $p = 0.036$). Current PwrAF users who recalled sound knee pain $\geq$4 NPRS with a passive foot reported significant and clinically meaningful improvements in sound knee pain (−2.5 NPRS, $p = 0.038$) and amputated side knee pain (−3 NPRS, $p = 0.042$). Current PwrAF users also reported significant and clinically meaningful improvements in patient-reported mobility (+4.6 points PLUS-M, $p = 0.016$). Individuals who had abandoned the PwrAF did not recall any differences between the feet.

Discussion: Current PwrAF users reported significant and clinically meaningful improvements in patient-reported prosthetic mobility as well as sound knee and

amputated side knee pain compared to recalled mobility and pain with passive feet used previously. However, a substantial proportion of individuals who had been fitted such a foot in the past did not recall improvements and had reverted to passive feet. The identification of individuals with unilateral TTA who are likely to benefit from a PwrAF remains a clinical challenge and requires further research.

Keywords: powered prosthetic ankle, powered prosthetic foot, powered prosthetic ankle-foot, knee pain, patient-reported mobility

INTRODUCTION

An amputation of a limb does not only remove passive anatomical structures but also results in the loss or truncation of muscles that are the actuators for movement and ambulation. Therefore, it appears consistent to develop powered prosthetic components that replace the function of the lost or impaired muscles. Thus far, however, passive components are still the standard of care in lower limb prosthetics. That requires individuals with amputations to adopt compensatory mechanisms to cope with the lack of power and active movement. In individuals with transtibial amputations (TTA), such compensations include slower walking speeds (1), about 25% higher energy expenditure for walking than able-bodied persons (2, 3), decreased sound limb step length (4), and reduced power generation in the residual knee (5). One important reason for these compensatory mechanisms is that passive prosthetic feet provide only up to 55% of the push-off power of the natural ankle-foot complex (6). Studies have shown that a commercially available powered prosthetic ankle-foot component (PwrAF) generates speed-dependent push-off power that may be comparable with that of the natural ankle (6–8). However, the results on its impact on function, such as self-selected walking speed (7, 9–13), metabolic energy expenditure on level ground (7, 9–11) and inclines (8, 10), patient-reported prosthetic function (12), and other aspects of prosthetic mobility have been inconclusive or conflicting.

Several studies have reported that walking with a PwrAF resulted in significant unloading of the knee joint of the sound limb (7, 13, 14). That is consistent with earlier findings that reduced push-off of the trailing limb requires increased collision work of the leading limb, which results in greater loading of its knee joint (15–17). This biomechanical evidence makes anecdotal reports from users of PwrAF on improved sound knee pain and pain-free walking distance noteworthy. Several studies with a PwrAF that did not find significant group benefits published the individual results of their subjects (9–11, 18). A thorough review of these subject-specific results revealed that, varying across the outcomes assessed, 35–50% of these individuals had experienced clinically meaningful benefits of the PwrAF during the studies. However, the published individual data has not allowed for narrowing down conclusive subject characteristics that would help guide the identification of individuals who are more likely to benefit from a PwrAF than others.

Therefore, it was decided to take a pragmatic, exploratory approach to systematically collect and analyze real-world, long-term user perspectives on musculoskeletal pain and prosthetic mobility in a bigger sample of individuals who were fitted a PwrAF in the past. The aim of the present study was to determine whether unsolicited anecdotal reports on reduced sound knee pain, amputated side knee pain, and low-back pain and as well as improved patient-reported prosthetic mobility were isolated occurrences or reflect a common experience in users of PwrAF. The results of the study, depending on the findings, were intended to serve as the basis for future planning of interventional studies.

METHODS

Study Design and Procedure

This was a pragmatic, exploratory cross-sectional clinical practice study approved by the Institutional Review Board (IRB) of the Baylor College of Medicine, Houston, TX, USA. A total of 250 individuals who had been fitted a PwrAF were invited to participate. All potential subjects had given prior written permission to contact them through email for research projects. The invitation email contained a link to the survey administered through Qualtrics® survey software. Subjects first provided IRB-approved informed consent and then answered questions on their demographics. Subjects completed the outcome measures for their current foot and provided the recalled ratings for their previous prosthetic foot. All responses were anonymous and de-identified.

The Device

Subjects had been fitted with one of two versions of a PwrAF, either the BiOM® T2 (BionX, Bedford, MA, USA) or the Empower® (Ottobock Healthcare LP, Austin, TX, USA). The BiOM was the earlier version and had been marketed from 2012 to 2017, whereas the Empower is the current version that has been available since 2018. Both devices have a combined ankle-motor/U-spring actuator mounted on an energy-storage-and-return (ESR) foot platform (**Figures 1A,B**) to provide actively powered plantarflexion/push-off during gait. The amount of push-off power delivered depends on patient weight, walking speed, terrain, and tuning of the software and reaches the level of able-bodied individuals (6–8). Compared to the previous BiOM, the current Empower was designed to have a more compact design without the protruding battery arm, achieve more consistent power delivery by more efficient springs and

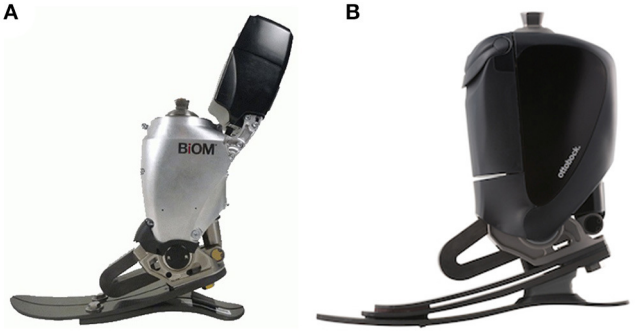

FIGURE 1 | (A) BiOM® powered ankle-foot component. **(B)** Empower® powered ankle-foot component.

improved tuning properties, and to extend battery life from a few hours to a full day.

Survey

The survey form inquired demographic information (age in 20-year bins, gender, height, weight), details on the amputation level and etiology, socket style and design, as well as prosthetic components. Subjects also completed the following outcome measures: numerical pain rating scales (NPRS) for average sound knee, amputated side knee, and low-back pain, the PROMIS Pain Interference Form-6a, and the Prosthetic Limb Users' Survey of Mobility–PLUS-M^{TM} 12-item Short Form. The online survey form is provided as **Supplementary Material**.

Outcome Measures

Numerical Pain Rating Scales (NPRS) are well-established and validated tools to assess pain on an 11-point scale from "0," representing no pain, to "10" representing the most intense pain imaginable (19–21). Subjects were asked, "How much pain do you suffer on average using your current prosthetic foot?" and "How much pain did you suffer on average using your previous prosthetic foot?" Pain ratings from 1 to 3 are considered "mild," 4 to 6 "moderate," and 7 to 10 "severe" (22). The minimal clinically important difference (MCID) has been reported to be 1 point or a 15% change (20). Improvements of 2 points or 30% have been found to correspond with a "much better" verbal rating of patients (23, 24). Numerical pain rating scales have been validated and used for remote electronic data collection (25, 26).

The PROMIS Pain Interference item banks assess self-reported consequences of pain on relevant aspects of subjects' lives including to what extent pain hinders engagement with social, cognitive, emotional, physical, and recreational activities (27). The pain interference short forms are universal rather than disease specific. They have been validated for diverse populations (28–30). In this study, subjects completed the 6-item original short form 6a. The response format was a 5-point ordinal rating scale of "Not at all," "A little bit," "Somewhat," "Quite a bit," and "Very much." Raw scores were converted to an item-response theory-based T-score using the PROMIS scoring manual (31). A T-score of 50 represents the average for the US general population with a standard deviation of 10 (27). A higher T-score indicates higher pain interference, and the MCID for T-score changes has been reported to be 2.0–3.0 (28, 29). The PROMIS Pain Interference has also been validated for remote (32) electronic data capture (30).

The PLUS-M is a validated and commonly used outcome tool based on a validated bank of 44 items to assess patient-reported mobility with a lower-limb prosthesis (33, 34). In this study, subjects completed the 12-item Mobility Short Form v1.1. The response format was a 5-point ordinal rating scale of "Unable to do," "With much difficulty," "With some difficulty," "With a little difficulty," and "Without any difficulty." Raw scores were converted to T-scores using the validated conversion table in the PLUS-M User Guide (35). Higher T-scores indicate better self-reported mobility. The minimal detectable change (MDC) has been reported to be 4.5 points (36). The PLUS-M has also been validated for electronic data collection (36).

Adjustment for Recall Bias

Patients are known to have a tendency to recall more pain but less functional limitations in retrospective postoperative assessments of preoperative knee and low-back pain and function compared to concurrent ratings prior to surgery (37–39). Therefore, an adjustment of the retrospective ratings was conducted to account for recall bias. Previous research comparing past concurrent and recalled ratings of pain and function reported an average recall error of 10% of the total range of the measurement tool (37). Thus, recalled pain ratings on the 0–10 NPRS were reduced by 1 point, except for original ratings of 0 or 1. For the PROMIS Pain Interference and the PLUS-M, the total raw scores for recalled ratings were reduced by 10% of their respective ranges, i.e., 2 points for the PROMIS Pain Interference (range 6–30) and 5 points for the PLUS-M (range 12–60), except if the raw score would have fallen below the minimum possible raw score. In that case, the raw score was adjusted to the minimum score. The recall-adjusted raw scores were then converted to T-scores as described above. However, adjustments for recall bias were not performed if they would have favored the PwrAF. Thus, recalled pain and pain interference ratings for the PwrAF in current passive foot (PAS) users as well as recalled PLUS-M ratings for PAS in current PwrAF users were not adjusted. We took this cautious methodological approach to reduce or possibly even prevent an overestimation of benefits of the PwrAF that subjects recalled and to prevent the creation of benefits that subjects did not recall.

Statistical Analyses

Descriptive statistics for the ordinal variables include the median, interquartile ranges (IQR), minimum and maximum values, and for T-scores means and standard deviations. Differences between the PwrAF and PAS scores were evaluated using Wilcoxon's matched pairs signed ranks test due to substantial departures from the normal distribution. This non-parametric statistical test is appropriate for the analysis of data where measurements of the same individual respondent are obtained under different conditions. McNemar's chi-square was used to test the significance of differences in proportions between

PWRAF and PAS. For all statistical tests, $p < 0.05$ were considered statistically significant.

Differences between PwrAF and PAS scores were analyzed for the two subject groups of current PwrAF users and current PAS users to avoid combining of current and recalled ratings for either foot in the statistical tests. In the group of current PwrAF users, current ratings for PwrAF were compared to the recalled original and adjusted ratings for PAS, whereas in the group of current PAS users, the current ratings for PAS were compared to the recalled original and adjusted ratings for PwrAF.

The individual results were analyzed in a descriptive way to find potential explanations for why each subject accepted or abandoned the PwrAF. Current and adjusted recalled ratings for the same outcome were analyzed for clinically meaningful differences between PwrAF and PAS. The three pain scores were summed up for a total pain score and differences ≥3 points NPRS were considered clinically meaningful. For the PROMIS Pain Interference and PLUS-M scores, differences ≥3.0 or ≥4.5 points were deemed clinically meaningful, respectively.

RESULTS

Demographics

A total of 52 individuals answered all questions of the online survey. Three subjects with bilateral transtibial and three subjects with transfemoral amputation were excluded. The responses of 46 individuals with unilateral TTA, all male, were subjected to the data analysis. This dataset represents a response rate of 18.4%. The demographic details of the subjects are depicted in **Table 1**. There were no significant differences between the groups of current PwrAF and current PAS users.

Eighteen subjects (39%) identified as current PwrAF users. Twenty-eight subjects (61%) reported to have used a PwrAF in the past but abandoned it because of its weight, limited battery life, or lack of waterproofness.

TABLE 1 | Demographics of the subjects.

	Entire sample	Current PwrAF users	Current PAS users
N	46	18	28
Sex male	46	18	28
Age			
20–39 years	8	2	6
40–59 years	23	9	9
60–79 years	14	6	8
80+ years	1	1	0
Height (cm)	181 ± 7	180 ± 6	182 ± 7
Weight (kg)	98.7 ± 15.4	100.2 ± 17.6	97.8 ± 13.7
Amputation etiology			
Trauma	37	16	21
Vascular disease	2	0	2
Cancer	1	1	0
Infection/Sepsis	4	1	3
Other	2	0	2
Time since amputation (years)	16.2 ± 11.3	19.1 ± 14.7	14.3 ± 7.8
Time of use of the PwrAF (years)	3.8 ± 3.0	6.6 ± 2.5	2.1 ± 1.8
Time of use of the PAS (years)	13.9 ± 10.9	16.0 ± 14.8	12.5 ± 6.9
Socket type with PwrAF			
Pin lock	14	4	10
Suction	17	9	8
Vacuum	11	4	7
Other	4	1	3
Socket type with PAS			
Pin lock	17	5	12
Suction	18	9	9
Vacuum	3	3	4
Other	1	1	3

TABLE 2 | Original pain and pain interference in current PwrAF users.

	PwrAF current ratings	PAS recalled ratings	*p*-value
Sound knee pain [median (IQR)]	1 (0–3)	2.5 (0.75–6)	0.007
Amputated side knee pain [median (IQR)]	1 (1–3)	3 (1–4.5)	0.007
Low-back pain [median (IQR)]	2 (1–5)	3.5 (1.75–6.5)	0.011
PROMIS pain interference [mean (±SD)]	50.9 (±7.4)	53.8 (±10.0)	0.173

IQR, interquartile range; SD, standard deviation.

Musculoskeletal Pain

All Subjects

In the original ratings, the 18 current PwrAF users reported significantly lower median current sound knee pain, amputated side knee pain, and low-back pain than they recalled for PAS. The difference in the PROMIS Pain Interference T-scores, though clinically meaningful in magnitude, did not reach statistical significance (**Table 2**). After adjustment of the recalled pain ratings for PAS for recall bias, only current sound knee pain remained significantly lower with PwrAF [1 (IQR 0–3) vs. 1.5 (IQR 0.75–5); $p = 0.036$] (**Figure 2**). The differences in amputated side knee pain [1 (IQR 1–3) vs. 2 (IQR 1–3.5); $p = 0.12$] and low-back pain [2 (IQR 1–5) vs. 2.5 (IQR 1–5.5); $p = 0.33$] were no longer statistically significant (**Figure 2**).

In the group of the 28 current PAS users, no statistically significant differences were seen between both foot types in the pain and pain interference ratings (**Table 3**).

Subjects Who Reported or Recalled Moderate to Severe Sound Knee Pain When Using PAS

After adjustment for recall bias, 13/46 subjects (28%; 6 PwrAF and 7 PAS users) reported current or recalled moderate to severe sound knee pain ≥4 NPRS with use of PAS, respectively. Most of these individuals also reported current or recalled amputated side knee and low-back pain at that level.

Current PwrAF users reported significantly and clinically meaningfully lower current median sound knee pain [3 (IQR 1.75–4.25) vs. 5.5 (IQR 5–7); $p = 0.038$] and amputated side knee pain [3 (IQR 1–3) vs. 6 (IQR 2.75–7), $p = 0.042$] than in the adjusted recalled ratings for their previous PAS. The differences in low-back pain [3 (IQR 0.75–5.6) vs. 7 (IQR 1.5–8); $p = 0.068$] and pain interference [54.5 ± 8.2 vs. 62.7 ± 4.2;

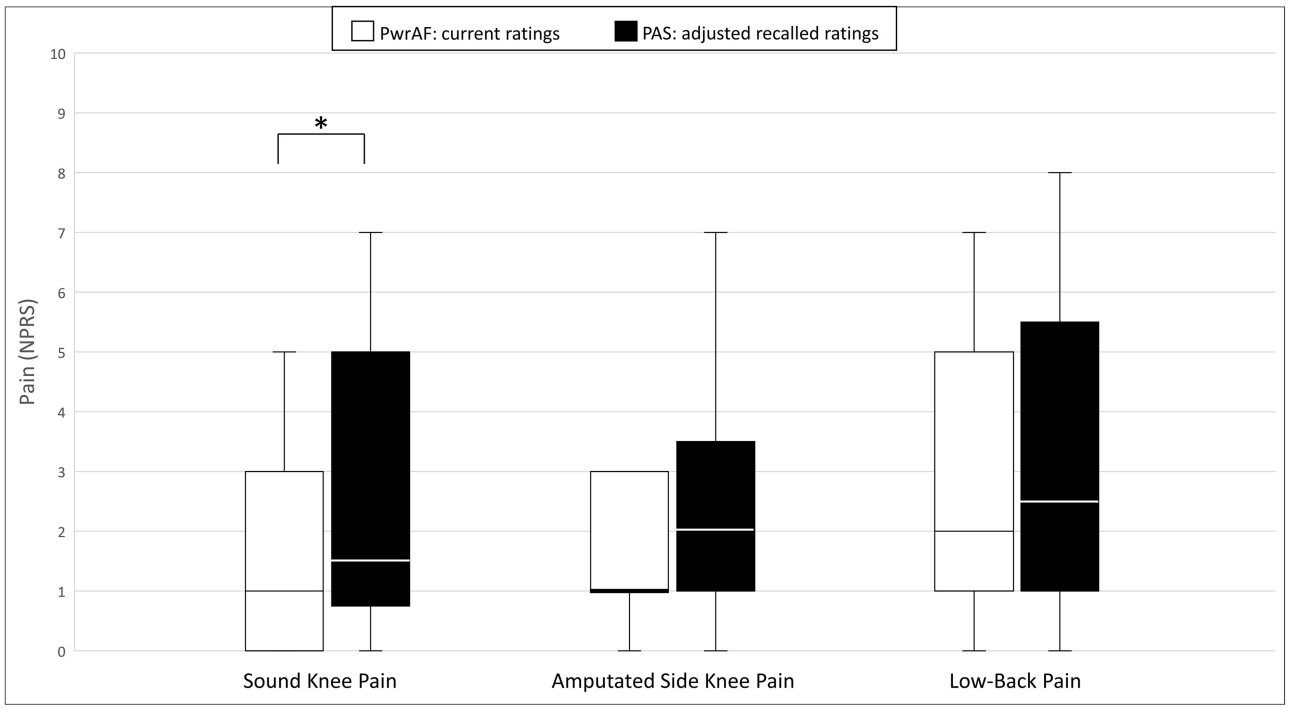

FIGURE 2 | Sound knee pain, amputated side knee pain, and low-back pain in current PwrAF users ($n = 18$). *$p < 0.05$ (see text for details). Differences in medians of 1 point or greater are considered clinically meaningful.

TABLE 3 | Original pain and pain interference in current PAS users.

	PAS current ratings	PwrAF recalled ratings	*p*-value
Sound knee pain [median (IQR)]	1.5 (0–3.75)	1 (0–3)	0.131
Amputated side knee pain [median (IQR)]	1 (0–3.75)	1 (0–3.75)	0.473
Low-back pain [median (IQR)]	2 (1–4)	2 (0.25–4)	0.823
PROMIS Pain Interference [mean (±SD)]	53.2 (±10.1)	53.1 (±9.6)	0.965

Recalled ratings for PwrAF were not adjusted (lowered) as this would have favored PwrAF. IQR, interquartile range; SD, standard deviation.

$p = 0.074$), though clinically meaningful in magnitude, failed to attain statistical significance in this small subgroup (**Figure 3**).

In contrast, the current PAS users who reported current sound knee pain ≥4 NPRS did not recall any significant differences in pain and pain interference between the feet (**Table 4**).

Patient-Reported Mobility

The group of 18 current PwrAF users reported significantly and clinically meaningfully higher current PLUS-M scores with PwrAF than they recalled for their previous PAS (54.9 ± 6.0 vs. 50.3 ± 7.8; $p = 0.016$) (**Figure 4**). No adjustment of the recalled PLUS-M ratings for PAS was performed as this would have further favored the PwrAF.

The 28 current PAS users did not recall a difference in PLUS-M scores with PwrAF to the current PLUS-M rating with PAS (52.4 ± 7.5 vs. 51.1 ± 7.8; p=0.071). The adjustment of the recalled PLUS-M ratings for PwrAF resulted even in a significant, though not clinically meaningful advantage for PAS (51.1 ± 7.8 vs. 47.4 ± 6.0; $p = 0.001$) (**Figure 4**).

The 6 current PwrAF users with sound knee pain ≥4 NPRS in the adjusted recalled ratings for PAS reported significantly and clinically meaningfully higher current PLUS-M scores with PwrAF than they recalled for their previous PAS (52.8 ± 3.9 vs. 40.8 ± 4.6; $p = 0.028$). In contrast, the 7 current PAS users with sound knee ≥4 NPRS pain did not report a significant difference in mobility between the feet, not even after adjustment of the PLUS-M ratings for PwrAF for recall bias (PwrAF 43.4 ± 8.5 vs. PAS 45.2 ± 11.4; $p = 0.735$) (**Figure 5**).

Group and Individual Outcomes by Version of the PwrAF

Of the 31 subjects who had been fitted the BiOM, only eight individuals (26%) were still using it at the time of the study. In contrast, 10 of the 15 individuals (67%) who had been fitted with an Empower were still current users.

The eight current BiOM users did not report any significant differences in pain, pain interference, and patient-reported mobility between PwrAF and PAS. After recall-adjustment, the 10 current Empower users reported significantly and clinically meaningfully lower current median sound knee pain [2.5 (IQR 1–3.25) vs. 4 (IQR 1–6.25); $p = 0.043$], amputated side knee pain [1 (IQR 1–3) vs. 2 (IQR 1–7); $p = 0.041$], and significantly and

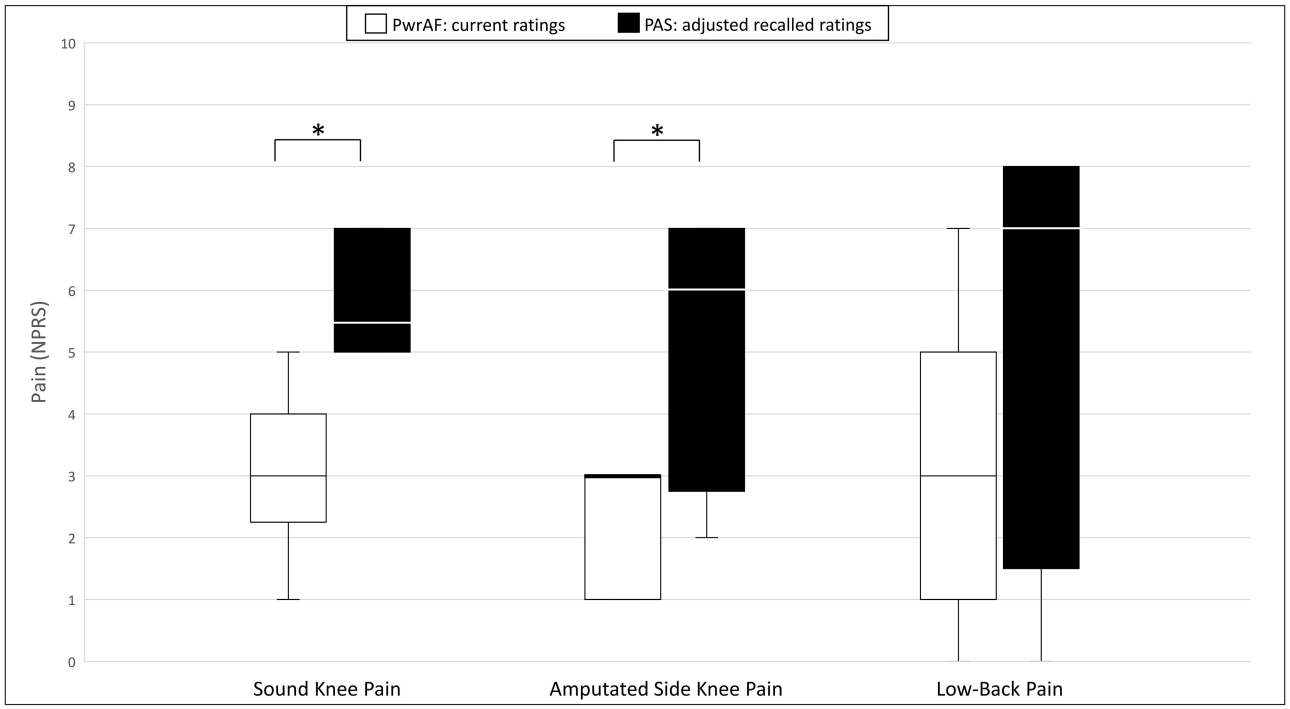

FIGURE 3 | Sound knee pain, amputated side knee pain, and low-back pain in current PwrAF users ($n = 18$) who recalled moderate to severe sound knee pain ≥4 NPRS when using PAS. *$p < 0.05$ (see text for details). Differences in medians of 1 point or greater are considered clinically meaningful.

TABLE 4 | Pain and pain interference in current PAS users who reported sound knee pain ≥4 NPRS when using PAS.

	PAS current ratings	PwrAF recalled ratings	*p*-value
Sound knee pain [median (IQR)]	5 (5–9)	5 (3–6)	0.063
Amputated side knee pain [median (IQR)]	4 (2–6)	4 (2–5)	0.465
Low-back pain [median (IQR)]	4 (3–7)	6 (2–7)	0.715
PROMIS Pain interference [mean (±SD)]	63.6 (±9.3)	61.9 (±8.5)	0.965

Recalled ratings for PwrAF were not adjusted (lowered) as this would have favored PwrAF. IQR, interquartile range; SD, standard deviation.

clinically meaningfully higher PLUS-M scores [55.1 ± 5.5 vs. 48.3 (±7.6); $p = 0.012$] than they recalled for the previous use of PAS.

On the individual level after adjustment for recall bias, 7 of the current 10 Empower users (70%) reported clinically meaningful improvements in the PLUS-M, and four subjects each (40%) in total pain and pain interference. Of the 8 current BiOM users, three individuals (37.5%) reported clinically meaningful improvements in the PLUS-M, and one subject each (12.5%) in total pain and pain interference. Summarizing the individual results, acceptance of the PwrAF may be explained by clinically meaningful improvements in the outcomes assessed in this study in 70% of current Empower and 37.5% of current BiOM users.

Likewise, abandonment of the PwrAF may be explained by the absence of clinically meaningful improvements in the outcomes used in this study in 5/5 (100%) former Empower and 17/23 (74%) former BIOM users. However, 6/23 individuals (26%) had abandoned the BiOM although they recalled clinically meaningful improvements when using the PwrAF.

DISCUSSION

The present pragmatic, exploratory study aimed at determining whether anecdotal reports of individuals with TTA on reduced musculoskeletal pain when using a PwrAF reflect a common experience among users or just isolated occurrences. As the identification of subjects who are likely to benefit from a PwrAF has been a clinical challenge (9–11), it was decided to survey a bigger sample of individuals who had been fitted such foot in the past to obtain real-world, long-term use experiences with the PwrAF and passive prosthetic feet. Though recall of ratings for previous interventions has limitations and challenges, this pragmatic approach is very similar to clinical practice where patients are usually asked to compare their current symptoms to those they recall for time points in the past. Recall of past symptoms and effects of sequential interventions is also the internal reference of the patient when a decision must to be made on the replacement of worn-out prosthetic components. However, as patients tend to overestimate past pain and function (37–39), recalled ratings were adjusted by clinically meaningful amounts according to recommendations in the literature (37), unless that would have favored the PwrAF. Therefore, it is reasonable to assume that statistically significant and/or clinically

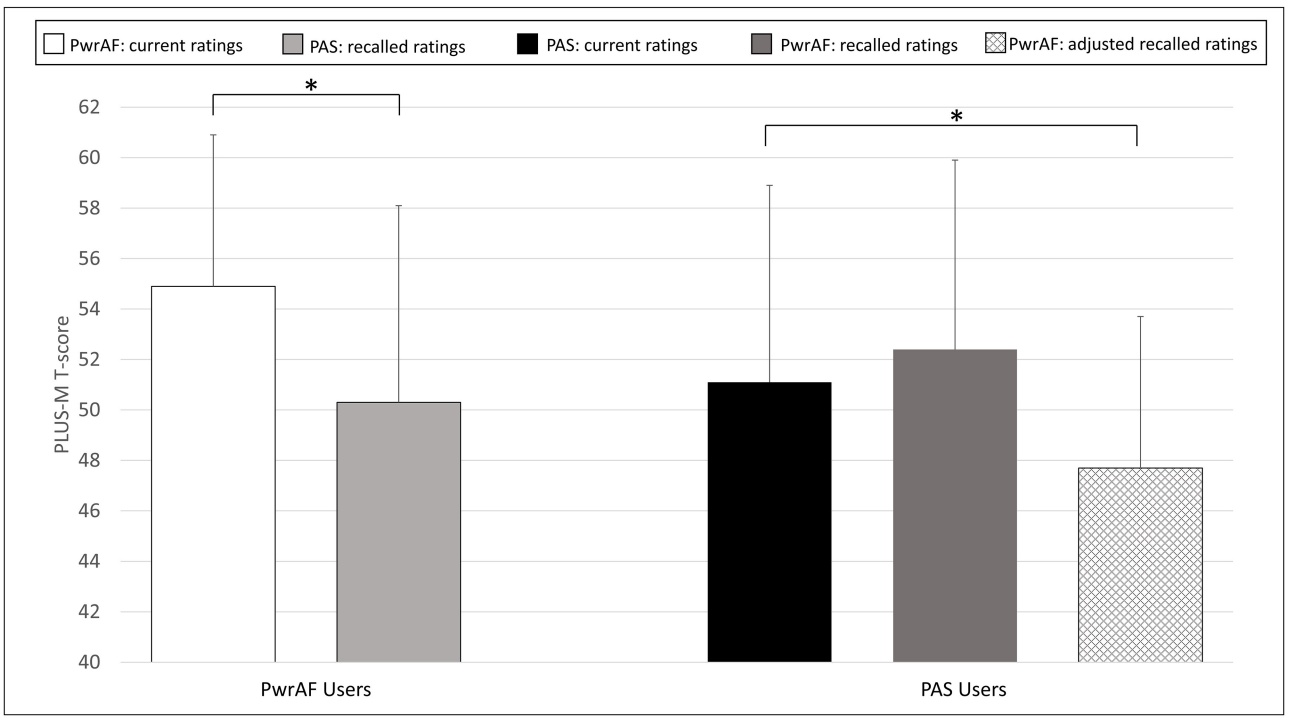

FIGURE 4 | Mean PLUS-M T-scores in current PwrAF users ($n = 18$) and current PAS users ($n = 28$). *$p < 0.05$ (see text for details). Differences in means of 4.5 points or greater are considered clinically meaningful.

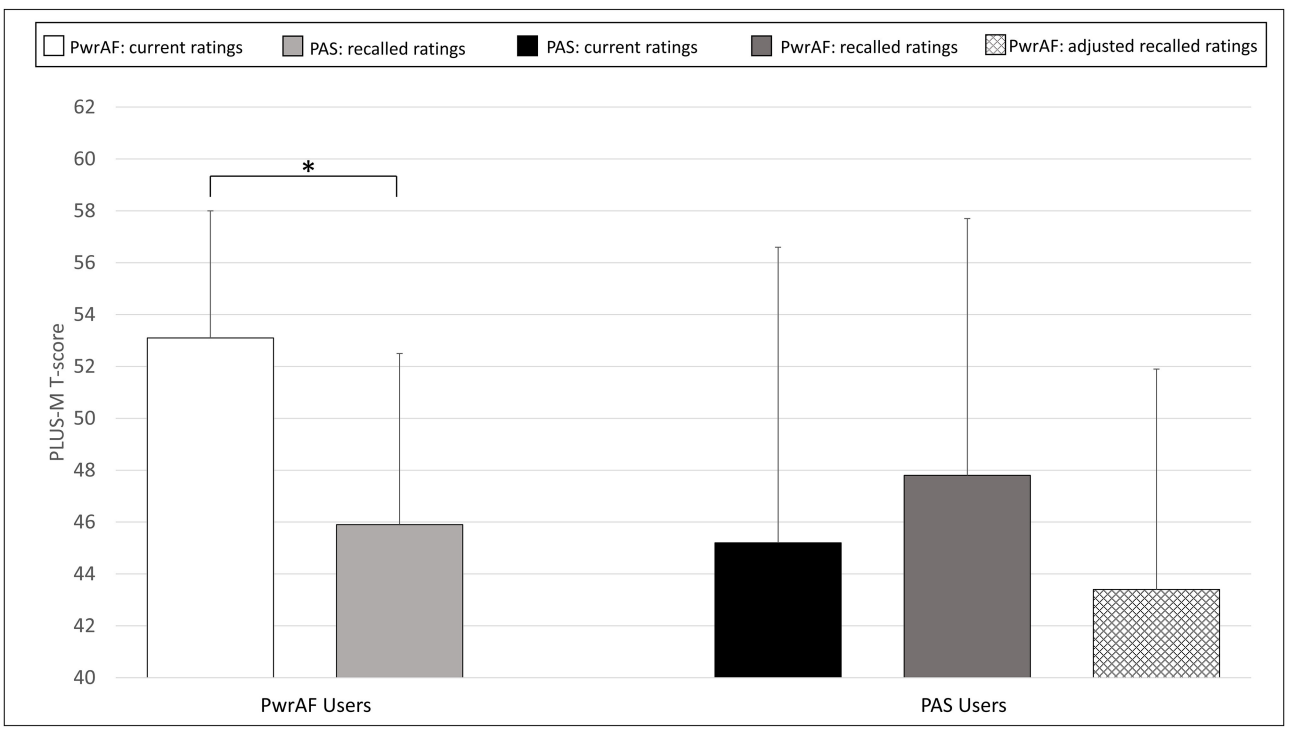

FIGURE 5 | Mean PLUS-M T-scores in current PwrAF users ($n = 6$) and current PAS users ($n = 7$) who reported sound knee pain ≥4 NPRS when using PAS. *$p < 0.05$ (see text for details). Differences in means of 4.5 points or greater are considered clinically meaningful.

meaningful differences that still exist after such substantial adjustments are worth contemplating and discussing.

Due to the limitations of recall, the discussion will take a conservative approach and focus only on differences between the two types of prosthetic feet after adjustment for recall bias. Current PwrAF users reported significantly less sound knee pain than for use of a passive foot. If current PwrAF users had recalled moderate or even severe sound knee pain ≥4 NPRS when using a passive foot, they reported statistically significant and even greater, clinical meaningful improvements in sound knee pain and amputated side knee pain. They also reported clinically meaningful improvements in low-back pain and pain interference that, however, did not reach statistical significance. To the best of our knowledge, there are no published studies that have investigated the impact of specific prosthetic components on musculoskeletal pain. However, there are publications that confirm that musculoskeletal pain is a clinical problem in individuals with lower-limb amputations (40–42). Our findings on pain reduction are consistent with published biomechanical mechanisms for unloading the sound knee, residual knee, and lumbar region by increased push-off power and/or increased passive ankle range of motion in a prosthetic foot.

Biomechanical Mechanisms for Unloading of the Sound Knee

For sound knee pain, it is important that push-off of the trailing limb produces forward and upward acceleration of the body's center of mass (COM) during step-to-step transition and reduces the collision work of the leading (sound) limb during landing (15–17). If the trailing prosthetic limb performs insufficient work during late stance phase to move the COM, the leading sound limb collides with the ground at a faster and downward directed speed (15), resulting in increased negative (eccentric) work to be performed by the sound limb's muscles and absorbed by its soft tissues and joints (17). Therefore, reduction of the negative work performed by the sound limb during collision may help reduce knee joint loading and the risk of developing knee osteoarthritis (43). The PwrAF investigated in this study has been shown to generate push-off that is comparable with that of able-bodied individuals (6–8). Consequently, biomechanical studies have found that the external knee adduction moment and other indicators of sound knee loading were reduced as compared to walking with a standard ESR foot (7, 13, 14). Though these reductions in knee loading were only statistically significant at faster walking speeds of 1.5 and 1.75 m/s (13, 14), they reached levels at medium walking speeds of 1.0 and 1.25 m/s that are considered effective for the treatment of knee osteoarthritis pain with knee unloader braces (44, 45). Thus, the reduction in sound knee pain found in this study may be explained by the biomechanical unloading of the knee of the sound limb associated with use of the PwrAF.

Biomechanical Mechanisms for Unloading of the Amputated Side Knee

For the effect on amputated side knee pain, it is instrumental that the PwrAF mechanisms surveyed in this study have a plantarflexion range of motion of 22° that can be used passively for fast foot-flat during level walking and terrain adaptation. In contrast, most current prosthetic feet have no articulating components. It has been reported that prosthetic feet with a controlled ankle joint facilitate smoother rollover and faster progression of the center of pressure while diminishing or even eliminating the "dead spot" phenomenon that is caused by an inappropriate recoil of the heel spring at about 20% of stance phase. All these effects of a prosthetic ankle joint result in decreased loading of the residual knee (46). In addition, the adaptability of the prosthetic ankle on slopes and uneven terrain reduces biomechanical compensations on the prosthetic and sound limbs, facilitates faster foot-flat, improves the control of downhill walking speed, and significantly reduces the biomechanical loading of the residual knee (47–50). These effects that have been shown for passive feet with non-microprocessor and microprocessor-controlled ankles may also be assumed for the powered ankle-foot components surveyed and may explain the reduction in amputated side knee pain.

Biomechanical Mechanisms for Unloading the Lumbar Spine and Muscles

For low-back pain, it is important that the loss of force and moment-generating capacity on the prosthetic side requires that the proximal muscles of the pelvis, hip, and lumbar spine participate in compensatory strategies to maintain balance and produce functional gait (51). These strategies often include complex recruitment of trunk muscles, co-activation of antagonistic muscles during stance, and asymmetric trunk posture at toe-off. While they support propulsion, they also result in high mechanical loads to the spine (51). Axial rotation of the lumbar spine is also increased during double-limb support, which may be a consequence of asymmetric trunk muscle strength and recruitment between the two legs (52, 53). These kinematic alterations in individuals with lower-limb amputation result in larger loads, loading rates, and load shifts compared to able-bodied individuals and are important risk factors that contribute to the onset of low-back pain (51). Increased prosthetic push-off has been shown to allow for better gait propulsion and force dissipation along the kinetic chain, thus reducing mechanical forces on proximal joints such as the knee, hip, and lumbar vertebrae (51, 54). As the prosthetic push-off produced by the powered ankles surveyed in this study reaches the natural push-off of able-bodied individuals (6–8), it may reduce asymmetries in pelvic and trunk muscle activation and improve force dissipation about the lumbar spine, thus alleviating low-back pain.

Current PwrAF Users Reported Increased Prosthetic Mobility

In addition to the reduction in musculoskeletal pain, current PwrAF users also reported a statistically significant and clinically meaningful increase in patient-reported mobility. This improvement was also significant and even twice as big in subjects who had recalled moderate to severe sound knee pain when using a passive foot. Only one earlier study had investigated patient-reported mobility with a PwrAF and did not

find significant differences to passive feet in their sample (12). Possible explanations for the improvement in patient-reported mobility with PwrAF use are the support of propulsion and ambulation by powered push-off and, in subjects with sound knee pain with PAS use, a reduction in musculoskeletal pain.

Why Don't All Individuals With TTA Benefit From a PwrAF?

As impressive as the results for pain reduction and patient-reported mobility in current users of a PwrAF are, it needs to be highlighted that a substantial proportion of individuals in our study had reverted to a passive foot and did not recall any differences during the time when they had used a PwrAF. On the individual level, even among the current PwrAF users, only 55% of subjects reported clinically meaningful benefits in patient-reported mobility and/or pain. Our findings are consistent with other studies that found either no or only limited benefits of a PwrAF in their entire samples, but detailed results showed that about 35–50% of their subjects had benefitted individually in walking speed, metabolic energy consumption, daily activity, or other aspects of prosthetic mobility (9–11).

That raises the question why only some persons with TTA appear to benefit from using a PwrAF. A systematic review of studies on prosthetic push-off power found that powered push-off had its greatest effects at walking speeds of $\geq$1.22 m/s, suggesting that individuals with high physical capabilities might be more likely to benefit from a PwrAF (6, 9). However, a later study found that 40–50% of its subjects who walked at slower speeds were also able to benefit from the PwrAF in self-selected in-lab walking speed and/or cost of transport (11). Another plausible explanation has been suggested by Kim et al. who studied muscle activation patterns when using a PwrAF (55). In able-bodied persons, push-off and propulsion are mainly driven by the gastrocnemius muscle (53), whereas current PwrAF act like the soleus muscle (10, 55) that has its main function in standing and postural control (56). As the external power is transferred differently than in able-bodied individuals (to the socket and residual limb below the knee vs. to the femur) and not integrated in the neuromuscular control of the user, subjects with TTA have to alter their neuromuscular control strategy to react to the added power and utilize it for propulsion (57). However, the study of Kim et al. did not find consistent muscle activation patterns in subjects with TTA while walking with a PwrAF, not even in long-term users (52). That suggests that motor learning and adaptation of neuromuscular control to the PwrAF may not be intuitive. Without a dedicated training program, some individuals with good motor learning skills may learn it fast, some may need a longer time, but a substantial portion of subjects may never learn to master it on their own. This is supported by our finding and that of others that only about half of the subjects benefitted individually and a few individuals even recalled more pain when using the PwrAF. Therefore, the current evidence suggests that the development of a specific gait training and rehabilitation program may help increase the proportion of individuals with TTA who could benefit from a PwrAF. For example, increasing the activity of the residual limb rectoris femoris muscle during walking may be a good strategy to stabilize the residual knee against flexion and, as a result, utilize the external ankle power more effectively by facilitating its transfer to the femur. In addition, activity of the gluteus medius and the medial hamstring on the amputated side appear to be correlated with the metabolic cost for walking (55).

Differences Between the Current and Previous Versions of the PwrAF

There were notable differences between the current Empower and the previous BiOM versions of the PwrAF. The proportion of current users was much higher for the Empower (67%) than for the BiOM (26%), and current Empower users reported significantly higher patient-reported mobility as well as significantly less sound knee and amputated side knee pain than with use of passive feet, while current BiOM users did not. Individuals who had abandoned either version reported no differences in the outcomes between PwrAF and passive feet.

On the individual level, acceptance of the Empower could be explained by clinically meaningful improvements in 70% of current users, while this was the case in only 37.5% of current BiOM users. Interestingly, more subjects experienced clinically meaningful improvements in patient-reported mobility than in pain. Abandonment of either PwrAF could be explained by the absence of individual clinically meaningful benefits in 79% of subjects. However, 21% of passive foot users, all former BiOM users, abandoned the PwrAF although they had recalled clinically meaningful benefits with its use. The likely reason for that is that the drawbacks of the BiOM technology, such as higher weight and limited battery life, outweighed the benefits for these individuals. Previous studies with a PwrAF only investigated short-term benefits and preference of the technology and did not report long-term benefits or device acceptance (7–14, 58).

There may be three possible explanations for the differences in clinical benefits and acceptance between the Empower and the BiOM. First, users of the Empower have been exposed to the technology for a much shorter period than users of the BiOM. Benefits may wear off or become less important and drawbacks more bothersome over time, especially as individuals age and decline in physical capacity. Second, technological improvements in the Empower in tuning, springs, and consistency of power delivery may have improved the ease of adapting neuromuscular control compared to the BiOM. Third, it cannot be ruled out that the sample of current Empower users consisted, by chance, of a greater number of individuals with excellent motor learning skills who had mastered the adaptation of their neuromuscular control and were therefore able to utilize the external power effectively. Further research is needed to identify patient characteristics and factors in the technology and rehabilitation program that help increase the number of responders who benefit from PwrAF.

LIMITATIONS

This study has limitations. First, it used recall for pain and prosthetic mobility for prosthetic feet that subjects had used in the past and compared them to ratings for the currently used

prosthetic foot. Subjects are known to tend to overrate past pain and physical function as compared to current ratings taken in the past (37–39). The risk of recall bias in this study was addressed by adjusting the recalled ratings by clinically meaningful amounts following recommendations in the literature (37). In addition, these recall adjustments were only performed if they resulted in a disadvantage for the PwrAF by narrowing the differences to the passive feet. A second limitation is the current inability to define predictive characteristics of responders to the PwrAF. Thus, our study could only survey a sample whose majority had not benefitted from using the PwrAF. A third limitation of this study is that no information was available on other factors that may have had an impact on musculoskeletal pain associated with prosthesis use, such as prosthetic alignment or concurrent medical treatments, such as physical therapy. Future research should assess musculoskeletal pain prospectively with current ratings of the studied devices and consider potential confounding factors. Third, of the 250 potential subjects asked to participate in the survey, only 52 (20.8%) responded, all of them male and 80% with traumatic amputations. It is unknown whether the results are representative for the entire population and may also be transferable to female individuals, subjects with other amputation etiologies, or whether the sample was overly skewed toward individuals who did not benefit from the PwrAF.

CONCLUSIONS

Free-living current users of powered prosthetic ankle-foot components reported significant and clinically meaningful improvements in patient-reported prosthetic mobility as well as sound knee and amputated side knee pain compared to recalled mobility and pain with passive feet used previously. However, a substantial proportion of individuals who had been fitted such a foot in the past did not recall improvements and had reverted to the use of passive feet. The rates of long-term acceptance and clinically meaningful benefits of the PwrAF device were much higher with the current than with the previous version. The identification of individuals with unilateral TTA who are likely to benefit from a PwrAF remains a clinical challenge and requires further research efforts. Nevertheless, a PwrAF is an option in the arsenal of the prosthetist and may be considered for individuals with unilateral TTA who suffer from musculoskeletal pain while using a passive prosthetic foot.

DATA AVAILABILITY STATEMENT

The raw data supporting the conclusions of this article will be made available by the authors, without undue reservation.

ETHICS STATEMENT

The studies involving human participants were reviewed and approved by Institutional Review Board of the Baylor College of Medicine, Houston, TX, USA. The patients/participants provided their written informed consent to participate in this study.

AUTHOR CONTRIBUTIONS

AK had the study idea, contributed to the study design, data analysis, and wrote the sections Introduction and Discussion. AM contributed to the study design, developed the survey, performed data collection, contributed to the data analysis, and wrote the sections Methods and Results. KH performed the statistical analyses. All authors contributed to the article and approved the submitted version.

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SUPPLEMENTARY MATERIAL

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Supplementary Data Sheet 1 | Online survey form.

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The remaining author declares that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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Visualizing the Unseen: Illustrating and Documenting Phantom Limb Sensations and Phantom Limb Pain With C.A.L.A.

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Currently, there is neither a standardized mode for the documentation of phantom sensations and phantom limb pain, nor for their visualization as perceived by patients. We have therefore created a tool that allows for both, as well as for the quantification of the patient's visible and invisible body image. A first version provides the principal functions: (1) Adapting a 3D avatar for self-identification of the patient; (2) modeling the shape of the phantom limb; (3) adjusting the position of the phantom limb; (4) drawing pain and cramps directly onto the avatar; and (5) quantifying their respective intensities. Our tool (C.A.L.A.) was evaluated with 33 occupational therapists, physiotherapists, and other medical staff. Participants were presented with two cases in which the appearance and the position of the phantom had to be modeled and pain and cramps had to be drawn. The usability of the software was evaluated using the System Usability Scale and its functional range was evaluated using a self-developed questionnaire and semi-structured interview. In addition, our tool was evaluated on 22 patients with limb amputations. For each patient, body image as well as phantom sensation and pain were modeled to evaluate the software's functional scope. The accuracy of the created body image was evaluated using a self-developed questionnaire and semi-structured interview. Additionally, pain sensation was assessed using the SF-McGill Pain Questionnaire. The System Usability Scale reached a level of 81%, indicating high usability. Observing the participants, though, identified several operational difficulties. While the provided functions were considered useful by most participants, the semi-structured interviews revealed the need for an improved pain documentation component. In conclusion, our tool allows for an accurate visualization of phantom limbs and phantom limb sensations. It can be used as both a descriptive and quantitative documentation tool for analyzing and monitoring phantom limbs. Thus, it can help to bridge the gap between the therapist's conception and the patient's perception. Based on the collected requirements, an improved version with extended functionality will be developed.

Keywords: limb amputation, phantom limb sensation, phantom limb pain, body image visualization, altered body image, documentation methodology, digital assessment, software tool

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INTRODUCTION

After the amputation of a limb, up to 90% of the patients report a feeling of the missing body part still being present (1). This effect is known as phantom limb sensation (PLS) and ranges from the simple feeling of presence to the perception of a specific posture, shape, or involuntary movements of the amputated limb (2–4). Additionally to PLS, which is defined as any sensation except pain (3), 45–85% of all patients suffer from phantom limb pain (PLP), which can manifest itself as e.g., stabbing, burning, twisting, or cramping (5). The term "phantom pain syndrome" was coined by Weir Mitchell in 1871 (4) when the use of the word "phantom" was commonly used in the medical field to describe pseudo-diseases, which may have contributed to the fact that PLP was stigmatized as "imaginary" for a long time (6).

PLP usually manifests itself 24 h to 1 week after amputation and decreases in intensity and frequency over time in most patients (3). Especially in the distal areas of the missing limb, PLP as well as PLS generally persist the longest. Some patients suffer from this pain for decades (2, 7). The underlying mechanisms causing PLP and PLS are still discussed controversially. The current dominant theory is the cortical remapping theory, according to which the brain responds to the loss of a limb with the reorganization of somatosensory maps: cortical areas that have received sensory signals from the amputated limb begin to receive input from neighboring areas (2, 4). Another explanation is based on the concept of a "neuromatrix"—an internal representation of one's own body. After an amputation, this representation remains intact and no longer matches the actual body, thus causing pain. The absence of visual and sensitive feedback of the missing limb enhances this effect (8).

PLP, defined as painful sensation in the missing part of the limb, is to be distinguished from pain in the residual limb (9), and in particular from neuroma pain. Painful neuromas develop at the stump of the severed nerve due to misguided attempts of nerve regeneration and are one of the main causes of residual limb pain (4, 10). Physical stimulation of the neuroma in form of pressure or stress on the limb can increase PLP, and in the past, neuromas were considered to contribute to the development and maintenance of PLP. However, PLP does also occur in the absence of stump pain, and removal of a neuroma does not cause PLP to disappear (2, 3).

PLP is an elusive entity, which makes it hard to track the progress of these patients over the course of treatment. Currently, there is no standardized mode of documenting PLP and PLS. The guidelines of the German Society of Neurology for the diagnosis of neuropathic pain recommend to document the onset and duration, the temporal course, pain qualities, localization and intensity as well as factors triggering pain (11). In general, it has become common practice to survey phantom pain with pain questionnaires. For example, the McGill Pain Questionnaire (MPQ) (12) became a de facto standard for the qualitative characterization of PLP, which is reflected by the terminology used in the medical literature after 1975 (13). Other pain questionnaires, such as the Brief Pain Inventory (BPI) (14), allow for the localization of pain by marking the appropriate areas on a 2D body chart. However, this type of documentation has the disadvantage of being not very precise. Shaballout et al. showed that a digital solution for drawing pain can not only contribute to a better understanding of the pain situation for physicians, but also facilitate analysis and quantification (15). Further improvement in the precision of this approach could be achieved by drawing pain directly on a 3D model (16).

This still does not allow for the illustration of the patients' altered body image, in particular the phantom. Although several software tools do exist that can be used to illustrate an altered body image, these have been developed primarily in the context of eating disorders (17–19). Therefore, the specific representation of a phantom limb is not possible with this approach. Appropriate illustrations would require an artist guided by the patient or could be drawn by patients with the appropriate drawing or photo editing skills (20). However, this is costly and totally unfeasible in a clinical context. Furthermore, it does not allow for a quantifiable analysis.

Since we could not find any suitable software, we decided to develop such a tool ourselves. In the present study we describe the functionality of the first version of C.A.L.A. (Computer Assisted Limb Assessment) and the results of its evaluation with therapists and patients in terms of usability and functionality.

MATERIALS AND METHODS

C.A.L.A.

The basic idea of C.A.L.A. is the customization of a virtual human 3D avatar in such a way that it represents the patient's body image including their PLS. A prototype (21) and a first version of C.A.L.A. were created by modifying and expanding the Open Source software applications MakeHuman (22), a software tool for 3D character creation, and the 3D modeling software Blender (23). This first version provided a 3D avatar that could be freely rotated and viewed from all sides and the principal functions of C.A.L.A.: (1) General adjustment of the 3D Avatar; (2) altering the shape of the phantom limb; (3) positioning the phantom limb; (4) drawing pain and cramps; and (5) the quantification of the created body image.

The process of documenting a patient over the course of treatment was as follows: Initially, a basic model is created by adjusting the 3D avatar to fit the patient's (perceived) body dimensions. This model then serves as a baseline to be built on in the following sessions. Over the course of treatment, the phantom limb can then be adjusted in terms of deformation, position, and pain, thus visualizing the changes in perception by the patient.

These functions are explained in detail in the following:

Adjusting the 3D Avatar

To increase the patient's identification with the 3D avatar, we used some of the original functions provided by MakeHuman, which allow for the adjustment of the avatar in terms of gender, age, muscles, weight, and proportions. These adjustments have no further purpose in the documentation process apart from cosmetic ones. The avatar can additionally be clothed with underwear.

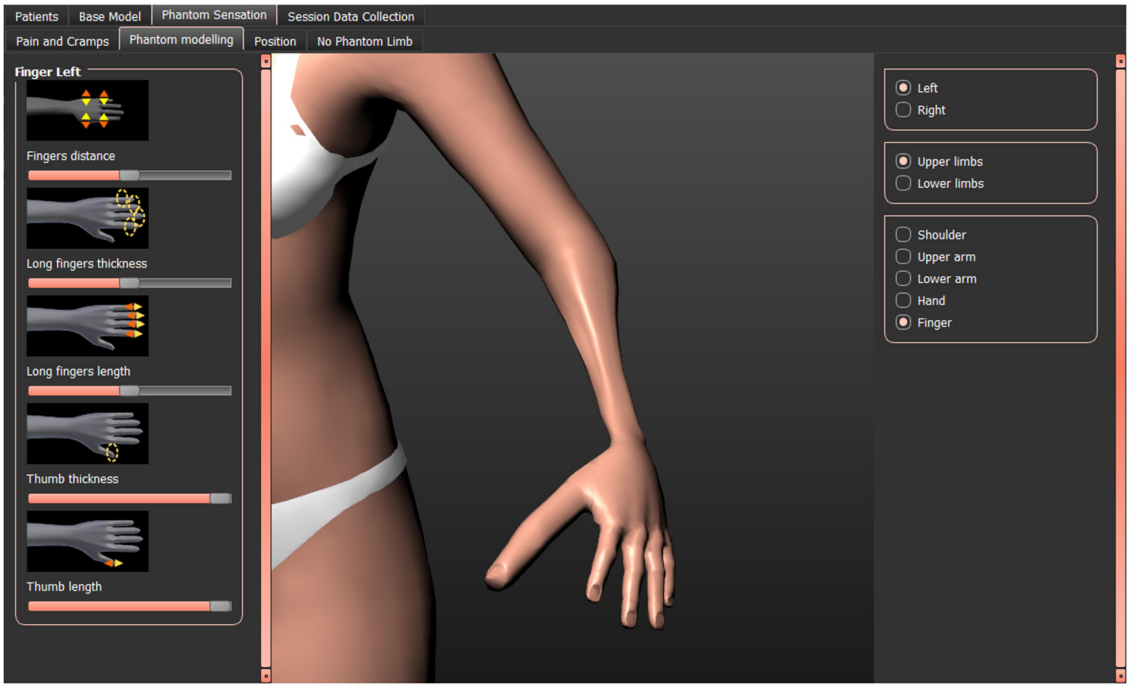

FIGURE 1 | Adjusting the shape of the phantom limb by decreasing the thickness of the lower arm and increasing length and thickness of the thumb.

Measuring the Patient

The patient's body measurements can be transferred to the avatar. The body height as well as circumference and length of upper arm, forearm, upper and lower leg, fingers, and toes as well as the length and width of hands and feet can be entered and form the basis for the subsequent measurements of the phantom limb.

Modeling the Phantom Limb

The length and circumference of the upper arm, forearm, thigh, and lower leg can be increased or decreased. Hands and feet can be enlarged or shrunk. Fingers and toes can be adjusted in length and circumference, the thumb and long fingers can be adjusted separately. The telescoping effect can be represented using this feature (see **Figure 1**).

Positioning the Phantom Limb

The sensation of the phantom limb being fixed in one or more, twisted or unnatural positions is captured by moving the respective joints of the 3D avatar into the position reported by the patient. Based on the original MakeHuman 3D model, it is possible to rotate the shoulder, elbow, and wrist joints as well as the individual finger joints of the 3D avatar, the same applies to the joints of the lower extremities. All joints can be rotated along their natural axes in steps of $\pm 10^{\circ}$ and even beyond the limits that are anatomically possible. As a result, all conceivable positions of the upper and lower extremities can be represented (see **Figure 2**).

Drawing Pain and Cramps

Pain is drawn directly onto the 3D avatar by using the mouse cursor as a brush, similar as it is done in 2D paint software. Currently C.A.L.A. distinguishes between pain in general and cramps in the phantom, these two aspects can be drawn independently of each other and with their respective intensity (see **Figure 3**), which is indicated by the Numeric Rating Scale (NRS) with a value between 0 and 10. The intensity is represented by different color schemes, general pain by a color gradient from yellow (slight pain) to dark red (severe pain), cramps by a color gradient from light blue (slight cramping) to dark blue (severe cramping).

Quantifying the Body Image

All data that were entered during the documentation process can be quantified and contain informative value about the phantom's constitution at the respective time. This allows for the analysis of the recorded aspects, namely deformation, position, and pain, and for their observation over the course of treatment. The quantification of these three aspects is briefly described as follows: Quantification of deformation reflects the percentage change in length and circumference of the respective limbs compared to the base model. Based on the originally collected dimensions of the patient's body, these changes can also be expressed absolutely in centimeters. The quantification of the position results from the deviation of each rotation axis of each joint from the basic position of the 3D avatar. Pain and cramps are quantified as the percentage of the body surface that is covered by the respective intensity.

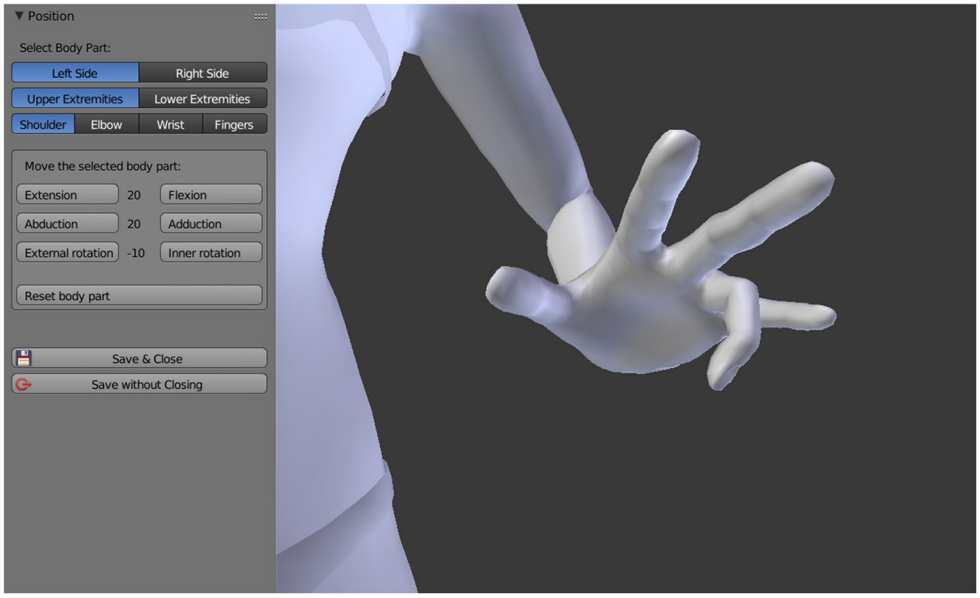

FIGURE 2 | Adjusting the position of the phantom limb by rotating the respective joint axes.

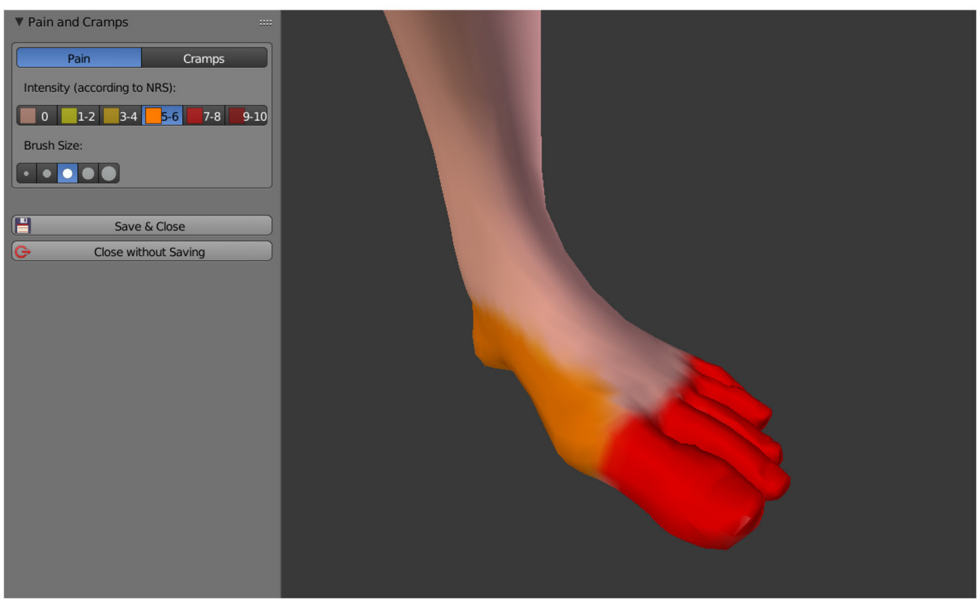

FIGURE 3 | Pain drawn directly onto the 3D avatar in different intensities.

Participants: Evaluation With Therapists

C.A.L.A. was evaluated with 33 professionals (19 physical therapists, 9 occupational therapists, 2 orthopedic technicians, 3 medical staff). Of these, 22 were female and 11 were male with an age range of 25–58 and a mean age of 41. The inclusion criteria for all participants were to actively work with amputees and document their phantom limb in a clinical context.

Each participant was initially provided with a brief introduction to the operation of C.A.L.A. Subsequently, participants were given the task to perform the entire documentation process (see Section C.A.L.A.) on two given, fictional patients (see **Supplementary Material**). These tasks were the same for all participants. It included the creation of a basic model, adjustment of the phantom's deformation, adjustment of the phantom's position, and

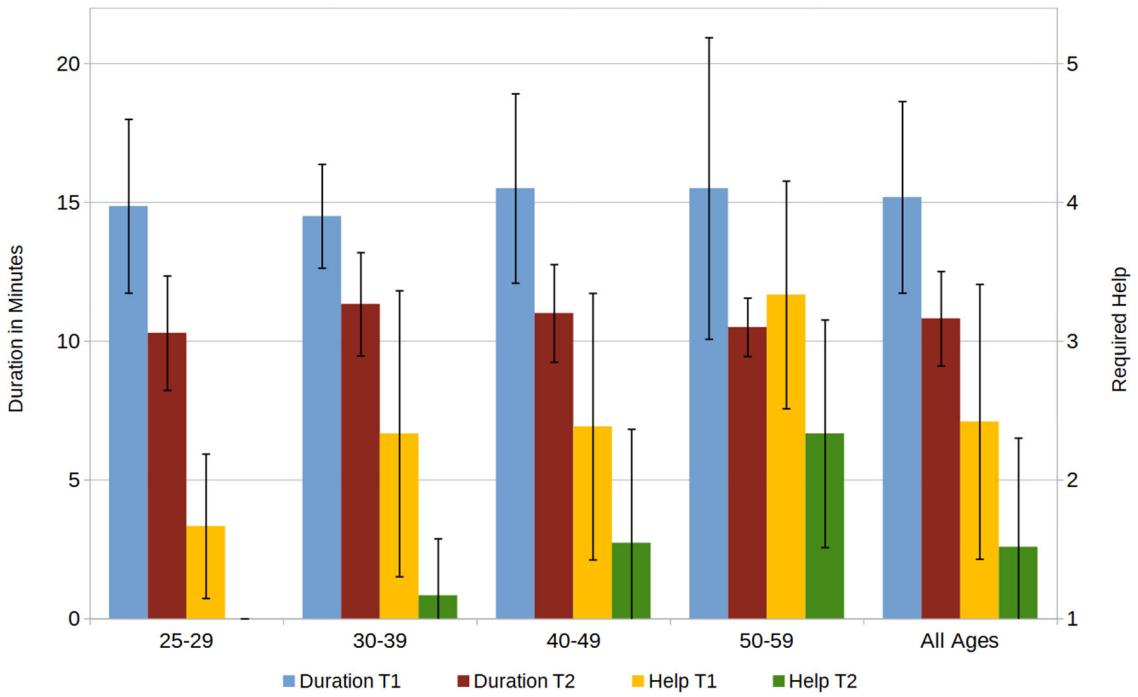

FIGURE 4 | Required duration for completion of the first and second task (T1, T2) in minutes and the required assistance, rated on a 1–5 Scale (1 = "no help," 5 = "a lot of help").

finally drawing pain and spasms. All participants were observed while performing the tasks and provided with assistance in operating the software. The duration for completing each task was measured and the level of assistance required was rated on a 1–5 Likert Scale by the investigator.

Subsequently, all participants were questioned with the System Usability Scale (24) to determine the user-friendliness of the software. With an additional self-developed questionnaire (see **Supplementary Material**) and semi-structured interview, the therapist's methods of documenting phantom pain and phantom sensation were surveyed and the principal functions of C.A.L.A. were rated. In the semi-structured interview, difficulties regarding the use of C.A.L.A., suggestions for improvement and additional desired functionalities as well as application scenarios were collected.

Participants: Evaluation on Patients

To test the scope of the currently implemented functionality regarding real-world cases of PLS and PLP, we evaluated C.A.L.A. on 22 patients with the following amputations: 1× transhumeral, 1× transradial, 12× transfemoral, and 5× transtibial, thereof one patient with a transfemoral and one with a transtibial amputation of both legs, 3× finger amputation. Eight of the patients were female and 14 were male with an age range of 21–73 and a mean age of 52. The inclusion criterion for all patients was the amputation of at least one limb.

For each patient, the entire C.A.L.A. documentation process was performed (see Section C.A.L.A.) by the investigators. The therapists who took part in our study did not evaluate the patients. Subsequently, the patients were questioned about their phantom pain with the German version of the Short Form McGill Pain Questionnaire (SF-MPQ-D) (25) to assess the presence of the different pain qualities. We administered a self-developed questionnaire (see **Supplementary Material**) with a 1–5 Likert scale rating system ("very inaccurate" to "very accurate") to determine how accurately the patients rated the representation of deformation, position, and pain of their phantom, and which aspects could not be mapped.

RESULTS

Evaluation With Therapists

All 33 participants completed the documentations of two given fictional patients. The average duration needed to complete a task decreased from 15.2 (±3.5) min for the first task (T1) to 10.8 (±1.7) min for the second (T2), the assistance provided by the investigator, measured on a 1–5 Likert scale ("very little help" to "very much help"), decreased from 2.4 (±1.0) to 1.5 (±0.8). Broken down by age group, the duration was very similar through all groups, however the amount of help provided was the highest for the oldest age group and the lowest for the youngest age group (see **Figure 4**).

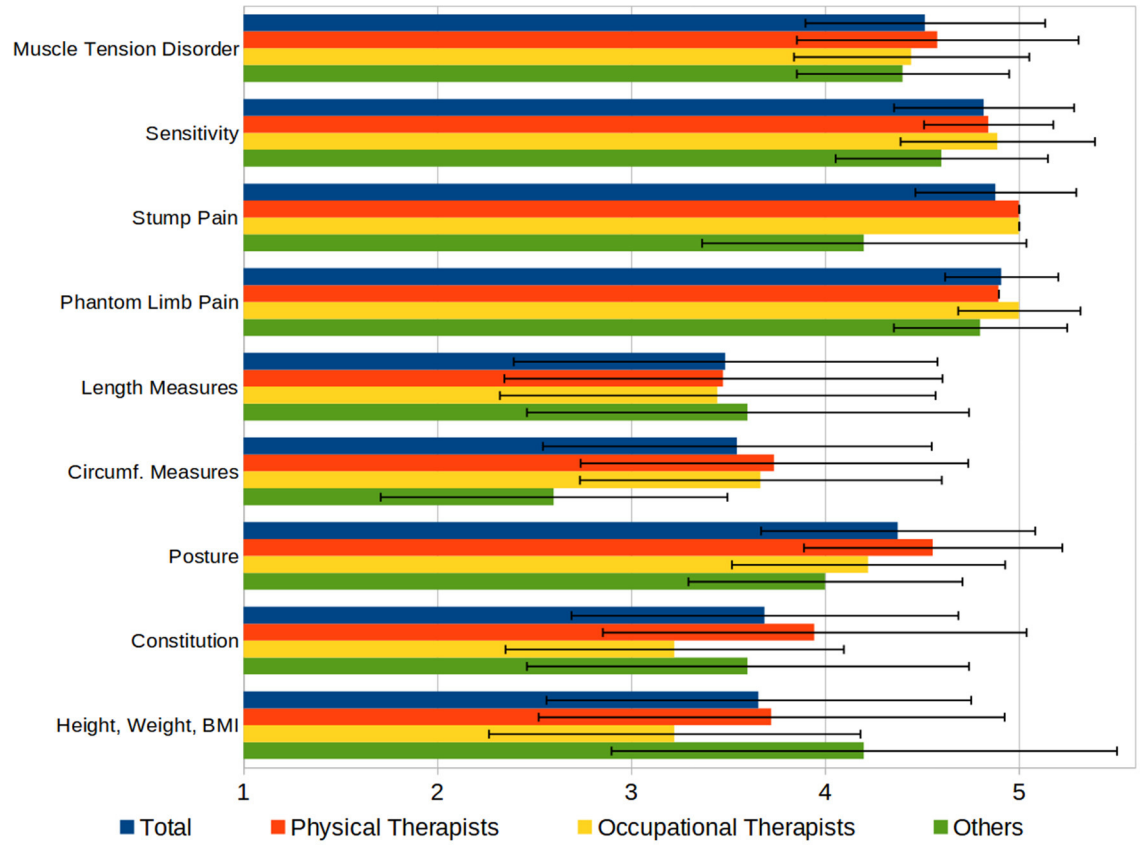

FIGURE 5 | Aspects of a therapeutic finding, rated by importance on a 1–5 Scale (1 = "not important," 5 = "very important").

The evaluation with the System Usability Scale resulted in an average score of 81.7% (±11.2), placing in the 4th quartile which represents high usability. The values are similar across age groups and professions. Additionally, we evaluated the usability of C.A.L.A. by user observation and semi-structured interviews, in which we asked about the difficulties in using C.A.L.A. Several users mentioned that the controls were too small and too cluttered. We also observed operational errors (such as modifying the wrong side of the body), problems understanding the user interface and difficulties navigating the 3D avatar.

With a separate, self-developed questionnaire and semi-structured interview we prompted the participants about their own documentation methods during therapy. Regarding the use of templates or specific questionnaires, 45% of the participants reported to use body charts to draw pain and 27% use validated questionnaires to assess pain, PLS, or body image. Besides questionnaires, the documentation was usually mostly handwritten and in a self-defined form.

We asked the participants to rate various aspects of the therapeutic finding by their importance on a scale from 1 to 5 ("not important" to "very important"). The results (see **Figure 5**) show the high importance of pain, sensation, and muscle tension, compared to the measures of the patient's body or their physical condition.

Regarding the documentation of pain, most of the participants (76%) used the Numeric Rating Scale (NRS) and 51% used the Visual Analog Scale (VAS) (26) to assess the intensity of pain. Twenty-four percent of the participants used pain questionnaires, with no questionnaire being reported more than once. Other aspects of pain such as influencing factors (e.g., medication, psychological state), temporal (24-h) course, and duration are documented in a free form. Questioned about the importance of several aspects of pain in documenting rated on a 1–5 scale ("not important" to "very important") showed that in average all aspects have been rated above 4.5 (see **Figure 6**).

Subsequently, we asked the participants to evaluate the functionality of the C.A.L.A. features. Using a Likert scale from 1 to 5 ("very low" to "very high"), participants were asked to rate the usability of the functionalities regarding the documentation of phantom limbs on a 1–5 scale ("not helpful" to "very helpful"), which resulted in high acceptance of the functions, rated least was the function to quantify the deformation of the phantom with 3.6 (±1.2) (see **Figure 7**).

In the semi-structured interview, we asked about additional functionalities for C.A.L.A., the most frequently mentioned ones are listed as follows: The documentation of pain qualities and the temporal aspects of pain (course, duration, frequency) were

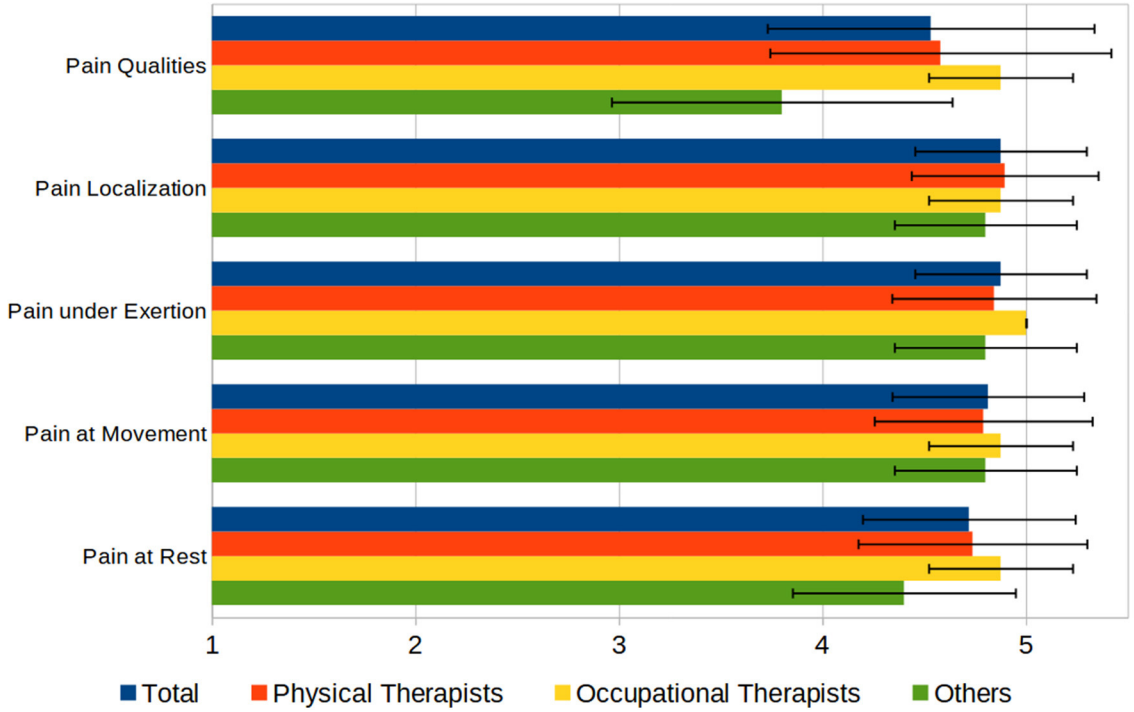

FIGURE 6 | Aspects of documenting pain, rated by importance on a 1–5 Scale (1 = "not important," 5 = "very important").

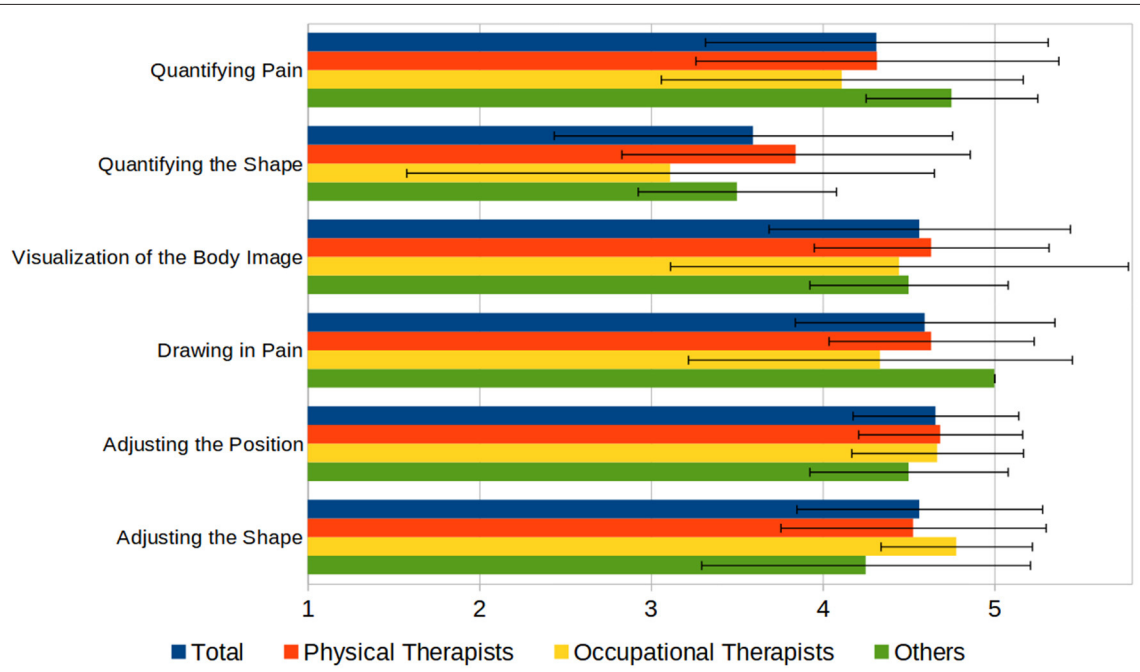

FIGURE 7 | Principal functions of C.A.L.A., rated by usefulness on a 1–5 scale (1 = "not useful," 5 = "very useful").

mentioned very frequently, not only in relation to phantom pain but also to residual limb pain. Another request concerned the modeling and positioning of the body parts, here a more differentiated adjustment, especially of the fingers and toes, was asked for. Regarding the positioning of the phantom, it was suggested to use a standardized diagnostic specification to

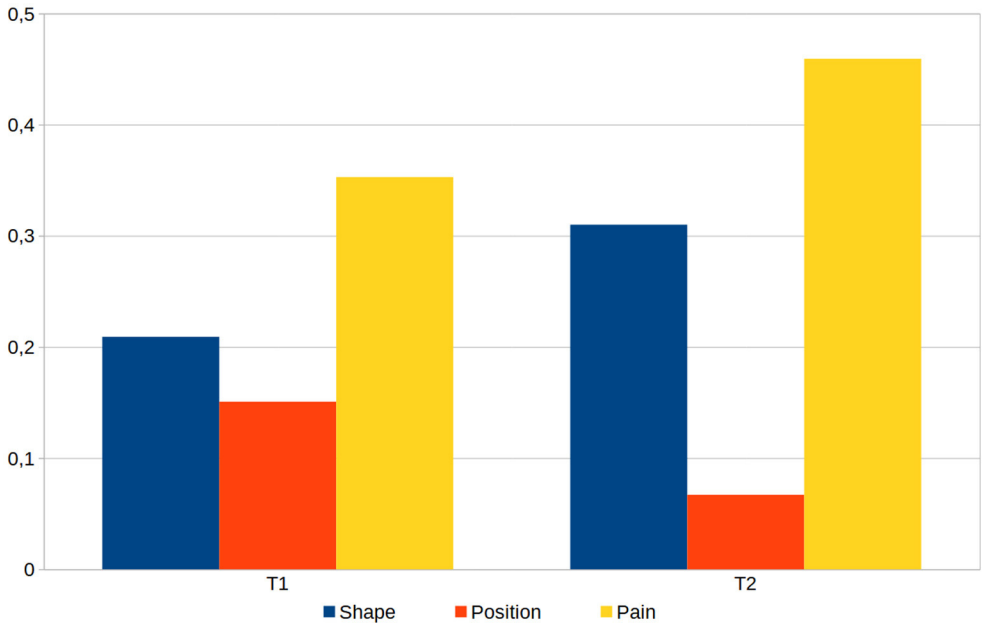

FIGURE 8 | The averaged coefficient of variance calculated for the three main aspects shape, position and pain, as determined from all 33 generated models for task 1 (T1) and task 2 (T2) of the therapist evaluation.

describe the rotation of the joints, e.g., the deviation from the neutral zero position (27).

Finally, we compared the quantifiable aspects of the documentations that have been created in the first respective second task by all participants, namely the shape and position of the phantom, and the drawing of pain and cramps. Analyzing the data revealed that during the documentation of both tasks, 3—always different—participants mixed up the side of the body and worked on the wrong arm or foot. The data of these six documentations were corrected by the side of the body and added to the evaluation. We determined the coefficient of variance for all parameters of the respective documentation aspects (deformation, position, and pain) and calculated their mean values (see **Figure 8**). This shows the smallest deviations when setting the position and the largest deviations when drawing pain, which is true for both tasks.

Evaluation on Patients

Seventeen of the 22 patients were experiencing PLP, five of them reported a deformed phantom and five reported a twisted position of the phantom. Thirteen patients reported suffering from stump pain.

All 17 patients with PLP or a deformed or twisted phantom were asked how well their phantom and their body image could be mapped, rated on a 1–5 Likert scale ("very inaccurate" to "very accurate"). The other five patients, who only reported stump pain were just asked about the accuracy of their body image representation. The phantom was rated with an average of 4.6 (±0.7), the body image with 4.2 (±1.0). We also asked the patients, how important the aspects of gender, age, and physical shape were for them regarding the accurate representation of body image, which revealed that these aspects were not of primary concern.

Regarding the functional range, it was often remarked that the body image was inaccurate due to the missing visualization of the residual limb. Modeling of the individual fingers was required in greater detail than provided, both in terms of deformation and position. It was also not possible to visualize, that some parts of the amputated limb were still present as phantom sensation while other parts were no longer perceived.

The documentation of pain revealed the missing option of documenting different qualities of pain. Here, especially the pain quality "stabbing" was mentioned several times. Another functional absence was the description of the temporal aspects of pain, such as long, short, or periodic pain. In addition, patients mentioned various other aspects when describing their pain, such as the course of the pain experience, the time of day, whether the patient was resting or moving, or even the influence of weather.

Subsequently, all 22 patients were interviewed with the SF-MPQ-D to measure number of pain qualities mentioned per patient and the frequency of each pain quality. For the patients without phantom pain, stump pain was queried instead. On average, 4.8 (±2.8) of 15 qualities were mentioned per patient, the most frequently mentioned were "shooting," "stabbing" and "hurting." During the interview as well as during the documentation of pain in C.A.L.A., it became apparent that the distinction between stump and phantom pain was not clear for many patients and therefore a mixture of both pain sensations was sometimes described.

DISCUSSION

Advantages of C.A.L.A.

The evaluation of C.A.L.A. with the System Usability Scale and the survey have shown that the vast majority of the participants considered C.A.L.A. user-friendly and feasible. To our knowledge, there is currently no software tool for therapists that allows for the visualization of phantom limbs, especially considering deformation, position, and pain. Therefore, rendering a comparison of C.A.L.A. to any existing standard regarding the documentation of phantom limbs is practically impossible. In a clinical setting time is of the essence. The duration of the documentation averages at 10 min, reducing the time by about one third in the second documentation trial, indicating that more training will likely reduce the time further.

During the evaluation with patients, they reported of never having given this amount of thought to the exact nature of their phantom limb. This fact was especially observable in localizing PLP. It was stated in only one case that the process of visualizing the phantom had a negative impact on the patient's body image. No patient indicated that phantom pain had increased because of the documentation process.

In this study we emphasized validity and did not specifically test for reliability, due to the nature of the modeling and positioning of the phantom limb and pain, which is dependent on the accuracy of the patient's report. We have provided different levels of detail in the tasks for position, deformation and pain, which is supported by the documentation differences shown in **Figure 8**. Especially regarding pain, room was intentionally left for interpretation, mimicking actual interactions with patients. In doing so, pain drawing could be assessed which resulted in the high variance. Corrective interventions during the dialogue with the patient could have lowered the outcome in variance.

Since all body image data are available in digital form, they can be easily quantified. This allows for a much more precise and simpler quantification than it would be possible with the conventional, mainly analog, methods. The amount of pain drawn onto a 2D human outline as well as joint angles of the phantom could possibly be estimated as could the circumference and the length. However, to our knowledge no one has ever calculated such values, especially regarding position and deformation, nor have their changes been evaluated over the course of treatment. To the best of our knowledge, no tools exist, yet, which can be used to document phantom limbs and PLP. C.A.L.A. offers a convenient tool to document just that.

In addition to evaluating usability, an essential aspect of our study was to identify possible extensions and adaptations of the functional scope. These will be discussed as follows.

The Struggle With Documenting Pain

When documenting PLP and PLS, pain is clearly the most important issue. Pain affects the patients' quality of life, and its reduction usually is the primary goal of therapy. The importance of pain was also evident in the qualitative surveys with patients, in which it was described by far the most frequently and in the greatest detail. In the therapist survey, too, there was the most feedback on the topic of pain documentation.

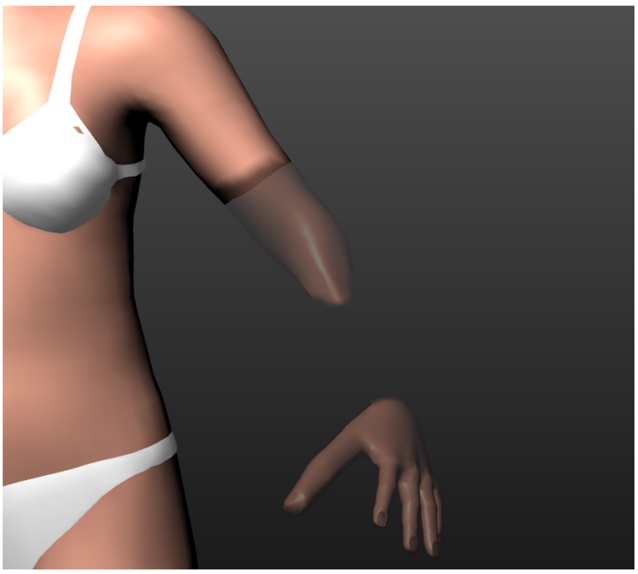

FIGURE 9 | A conceptual illustration of how to visualize the residual limb and phantom limb. The "presence" of the phantom limb is indicated by its visibility, meaning that the invisible parts are no longer perceived by the patient.

In this context, the topic of pain qualities was most frequently mentioned by both patients and therapists. This is not surprising since using these pain qualities for describing PLP had been established almost 50 years ago (13). Currently, in C.A.L.A. it is only possible to enter "general" pain and the pain quality "cramping." Expanding this to document other pain qualities seems useful, whereby clustering them to a few 5–10 qualities would be necessary. The current method of evaluating the pain intensity using the NRS is a common approach among the interviewed therapists (used by 76%).

In addition to the localization, intensity and the qualities of pain, the guidelines of the German Society of Neurology (11) recommend documenting the aspects of duration and temporal course as well as the factors that trigger pain. In addition, the qualitative evaluation also revealed quite a few other aspects of pain relevant for a complete description, e.g., deep/superficial pain. However, all mentioned aspects have in common that their visualization in C.A.L.A. would be difficult and not very intuitive to understand. We therefore consider it useful to omit these aspects from the documentation of PLP in C.A.L.A.

Representation of Phantom and Stump

Besides the issue of pain, C.A.L.A. should include means of clearly visualizing the stump to make it easier to distinguish it from the phantom. Several patients stated during the qualitative interviews that the visualization of their body image was not complete due to the missing visualization of the stump, even if the sensation of the phantom limb was present in the patients.

When drawing PLP based on the patients' descriptions, it became obvious that the strict distinction between phantom pain (exclusively in the missing part of the limb) and stump pain

(exclusively in the part still present) (9) was not necessarily useful for the patients. Although we pointed out that we intended to document only phantom pain, in some cases the pain described extended from the phantom to the existing limb, in a few cases even to the middle of the body.

The distinction between PLP and residual limb pain could be simplified in a future version of C.A.L.A. by a clear visual differentiation between residual limb and phantom limb in the representation of the 3D avatar (see **Figure 9**). This would make it more obvious, both when drawing and when evaluating pain, whether the pain is located at the stump or actual PLP is experienced. Considering that pain in general is probably the most important aspect of quality-of-life-limiting discomfort, we consider it useful to expand C.A.L.A. to include the documentation of stump pain as well.

As described in the literature (2, 20) and also observed in some patients, phantom sensation was not present in the entire lost limb, but only in the distal areas of the phantom. To increase the precision of the representation, this circumstance could be represented by masking the areas of the phantom that are no longer perceived (see **Figure 9**).

Adjusting the Functionality

In addition to these two main topics, we have identified several other contexts in which C.A.L.A. could be improved to increase its usability and validity. The most relevant ones are listed below.

When adjusting the position of the phantom limb, the 3D avatar is initially in a position where arms and fingers are slightly spread and bent. While this body position is advantageous for painting and deforming the phantom, we think that a more standardized body position, such as the neutral-zero position (27), would be more beneficial for phantom limb positioning. We believe that such an alignment of the initial position will not only facilitate the positioning of the phantom, but will also increase the significance of the quantified position. The range of functions concerning the positioning and deformation of the phantom has shown that the currently provided options can only partially cover the large variety of different perceptions. Especially for hand and fingers, but also for foot and toes, it would be required to allow adjusting them in further detail.

Another feature that has been mentioned several times was the desire for a visual representation of the progress of the phantom over the course of treatment; or, in other words, over the course of several documentations. This could especially help both to clearly demonstrate the progress of therapy and to motivate the patients to continue.

Finally, participants also considered other possible application scenarios in which C.A.L.A. could be used with modified functionality. Often mentioned was the application in Complex Regional Pain Syndrome (CRPS) or stroke patients, as well as for all other situations in which patients experience an altered body image.

CONCLUSIONS

We have created a tool that allows for the visualization and documentation of PLP and PLS. Thus, it provides a standardized form for their presentation and can be used as a descriptive and quantitative documentation method.

Based on the evaluation with the therapists, a great demand for our tool could be determined, therefore a further development of C.A.L.A. is reasonable and can contribute to increase its usability and efficiency in operation. For such an improved version, the most important additional features in our point of view are briefly listed here again: (1) introduction of pain qualities; (2) clear distinction between phantom and residual limb; (3) additional documentation of residual limb pain; (4) more precise adjustment of shape and position of individual fingers; and (5) a visualization of the course of treatment over several sessions.

C.A.L.A. can help to bridge the gap between the therapist's conception and the patient's perception of the phantom limb. The possibility to quantify the representation of the phantom offers a previously unavailable option to monitor and analyze its change over the course of treatment and can help to create insights into the correlation between certain forms of treatment and PLS or PLP. Finally, C.A.L.A. enables a more integrated representation of the phantom than is possible with conventional visualization methods with little effort regarding time and other resources, increasing feasibility regarding clinical context.

DATA AVAILABILITY STATEMENT

The raw data supporting the conclusions of this article will be made available by the authors, without undue reservation.

ETHICS STATEMENT

The studies involving human participants were reviewed and approved by Ethics Committee of the Medical Faculty of the Eberhard-Karls-University Tuebingen. The patients/participants provided their written informed consent to participate in this study.

AUTHOR CONTRIBUTIONS

MB designed and implemented the first version of C.A.L.A., conceived and performed the evaluation, analyzed the data, and wrote the manuscript. JM contributed to the design of C.A.L.A., contributed to planning and performing of the evaluation, and revised the manuscript. JH provided feedback on and verified the data analysis and revised the manuscript. MVB contributed to the interpretation of the results and revised the manuscript. AD and JK contributed to critical revision of the intellectual content and approved the final version. CP conceived the original idea of the project, contributed to the design of C.A.L.A., contributed to data acquisition, revised the manuscript, and was in charge of overall direction and planning. All authors provided critical feedback and helped shape the research, analysis, and manuscript. All authors contributed to the article and approved the submitted version.

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SUPPLEMENTARY MATERIAL

The Supplementary Material for this article can be found online at: https://www.frontiersin.org/articles/10.3389/fresc.2022.806114/full#supplementary-material

A supplementary video explaining and illustrating the functionality of C.A.L.A. can be found here: https://www.playbionic.org/cala/.

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Conflict of Interest: The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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Phantom Sensations Influenced by Global and Local Modifications of the Prosthetic Socket as a Potential Solution for Natural Somatosensory Feedback During Walking: A Preliminary Study of a Single Case

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Following lower limb amputation, amputees are trained to walk with a prosthesis. The loss of a lower limb deprives them of essential somatosensory information, which is one of the causes of the difficulties of walking with a prosthesis. We here explored whether a solution to this lack of somatosensory feedback could come from natural sensations of the phantom limb, present in most amputees, instead of from substitutive technologies. Indeed, it is known that phantom sensations can be modulated by (i) global mechanical characteristics of the prosthesis socket, and (ii) locally applying a stimulus on an area of the residual limb. The purpose of this pilot study was to verify the feasibility of influencing phantom sensations *via* such socket modifications in a participant with transfemoral amputation. Four prosthetic interface conditions were studied: a rigid and a semi-rigid socket, each one with and without a focal pressure increase on a specific area of the residual limb. The results show that phantom sensations during walking were different according to the 4 interface conditions. The participant had more vivid phantom sensations in his foot and calf of which some varied as a function of the gait phases. Preliminary gait analysis with wearable sensors shows that these modifications were accompanied by changes in some gait spatiotemporal parameters. This preliminary study of single case demonstrates that phantom sensations can be modulated by the prosthetic interface and can provide natural somatosensory information dynamically varying with gait phases. Although this needs to be confirmed for a larger population of lower limb amputees, it already encourages non-painful phantom sensations to be considered early during the rehabilitation of lower limb amputees.

Keywords: amputation, lower limb, referred phantom sensations, prosthetic socket, sensory feedback, gait, interface

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INTRODUCTION

Background

Walking is heavily based on somatosensory feedback that informs about the state of the body and its interactions with the environment (1). Indeed, skin afferents from the plantar sole (2–5) and muscle and joint proprioceptive information (6, 7) are crucial for maintaining balance. After lower limb amputation, one is forced to walk with a prosthesis, thus without most of this feedback. Walking, an almost automatic behavior in a non-amputee, becomes a tiring activity that requires significant cognitive resources (8–10). Moreover, this lack of somatosensory information could be one of the causes of the asymmetry (8, 11) characterizing the gait of amputees (12–14). Asymmetry is probably involved in the increased prevalence of degenerative pathologies of the contralateral limb (15–17). Therefore, the loss of lower limb somatosensory feedback impacts daily activities and could cause long-term health problems.

The provision of somatosensory information on the position of the prosthesis or ground characteristics has been found to improve the efficiency of prosthetic walking. Through implanted electrodes, a team of researchers stimulated the peripheral nerves of the residual limb to give feedback about pressure detected by sensors on the prosthetic plantar sole and about the flexion angle of the prosthetic knee (18). Many variables related to prosthetic gait efficiency (cognitive load, oxygen consumption, walking speed, number of falls) were improved by this feedback, which is promising. Other work has demonstrated the value of providing somatosensory information using a non-invasive technique by instrumentation of the prosthesis alone. For example, after a training period, cutaneous vibratory stimulation—when consistent with the phases of the prosthetic stance phase—reduced temporal gait asymmetry and increased stride length without increasing cognitive load (11). These studies suggest that additional information about gait phases leads to more efficient prosthetic walking. However, these studies used invasive or instrumented solutions that are non-intuitive and therefore need a training period. The present study consisted in exploring a new solution allowing intuitive somatosensory feedback during walking through naturally present phantom limb sensations.

A majority of amputees still perceive the lost limb through a natural phenomenon called "phantom limb" (19–21). These phantom sensations are often non-painful, manifesting as tingling, warmth, or a simple sense of presence as an intact limb (19, 22). Painful phantom sensations also exist but, contrary to popular belief, are not as common as non-painful sensations and are often occasional and moderate (21). Non-painful phantom sensations during dynamic and functional activities such as walking have only been little reported in the literature. Yet, according to our recent interviews with 97 lower limb amputees, 30% of the participants report perceiving their phantom limb when walking. Moreover, the phantom limb can not only be present during walking, but even interact with it because it were perceived as either indispensable, as an aid, or sometimes as a disturbance. This suggests that these natural sensations may compensate for the lack of somatosensory information if they are coherently varying with the gait phases. Interestingly, in some amputated persons, stimulation of the residual limb modulates phantom sensations. These sensations are called referred sensations (RS) (22, 23). One can distinguish local and global stimulation. Local stimulation [static or dynamic pressure (23, 24), vibration (25), or electrical stimulation (26)] applied directly on specific areas of the skin can evoke changes in phantom sensations in some amputees. Regarding global stimulation, we found that for many participants, wearing their prosthetic limb, even without seeing it, changed their phantom sensations. Thus, global pressure applied on the residual limb by contact with the prosthetic socket seems to be a form of stimulation that influences phantom sensations. This is interesting because local stimuli in the socket or global stimuli related to the socket itself could be exploited to induce RS that vary coherently with the gait phases and would thus be usable as a form of somatosensory feedback.

Objectives

The main objective of this preliminary study was therefore to investigate the feasibility of modifying the phantom sensations perceived during walking by intervening on the prosthetic interface of a lower limb amputee participant. To study whether the different interface conditions also influenced the gait pattern of the participant, a preliminary analysis of some gait parameters was performed.

METHODS

Study Design

This study is a single case cross-sectional study since we observed and analyzed the effect of an intervention on the prosthetic socket at a specific point in time.

Participant

The participant was 48 years old and had a traumatic transfemoral amputation of his left leg 5 years before the present study. He daily wore a prosthesis with an adherent semi-rigid socket without a sleeve, could walk long distances, and did not take medication. He reported neither phantom nor residual limb

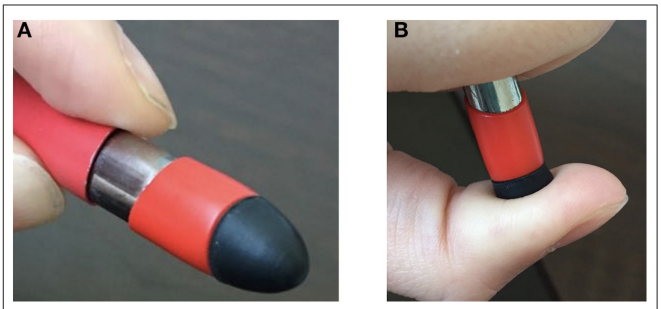

FIGURE 1 | The instrument used to explore referred sensations (RS) by stimulating different areas with the soft rubber tip **(A)**, pressing on the skin **(B)**. The participant reported RS during intense pressure.

pain, and he was able to describe his phantom sensations in detail. Written informed consent was obtained before the study.

Setting

This study was conducted in 4 phases, all in the Chantecler Rehabilitation Center (Marseille, France) and in the presence of medical staff. At the end of March 2021, a semi-structured interview regarding the participant's phantom sensations was conducted. Then, the participant was seen 3 times during April 2021, first to perform the mapping of RS, then to design the 4 prosthetic interface conditions, and finally, to make the participant walk with the 4 interfaces to analyze his phantom sensations during walking and some gait spatiotemporal parameters.

Phase 1: Exploration of Phantom Limb Sensations With a Semi-structured Interview

The semi-structured interview allowed for a detailed description of non-painful and painful phantom sensations as well as factors influencing them. Phantom sensations were described according to their nature (e.g., tingling, touch, heat, "muscular contraction") and their location (e.g., toes, heel, calf). Furthermore, we asked the patient to describe the factors influencing them (e.g., weather, fatigue, activity), and explored, in particular, the influence of different conditions: sitting with and without a prosthesis, standing with a prosthesis, and while walking. The interview was completed by a clinical examination to search for RS by palpation of the residual limb. The interview lasted 1.5 h.

Phase 2: Mapping of Referred Sensations

As the interview showed that the participant had RS, a detailed mapping was carried out to identify both the areas of the residual limb for which stimulation could modify phantom sensations and the nature of the RS. Intense localized pressure on the skin appeared to induce RS in the phantom limb (**Figure 1**). A total of 69 areas of the residual limb were defined by dividing the residual limb into 4 faces (medial, lateral, posterior, and anterior), each divided into 4 horizontal strips of 5 cm high, each comprising 6 areas to be stimulated (**Figure 2**). The participant was standing during the mapping.

Phase 3: Global and Local Modifications of the Prosthetic Interface

Based on the RS map, a patch was designed and placed inside the socket to apply pressure on skin areas that induce RS potentially useful for walking, thereby locally modifying the prosthetic interface (**Figure 3**). Furthermore, to study the influence of global changes of the prosthetic interface, two types of sockets made of different materials were used: rigid plastic called glycolic polyester (PETG) and semi-rigid plastic called ThermoLyn® that is surrounded by carbon support. The form and dimensions of these two sockets were the same, as were the alignments of all prosthetic components. The rigid socket had already been worn daily by the participant as it had been his temporary socket.

Phase 4: Description of the Phantom Limb Sensations When Walking and Recording of Gait Parameters for Each of the 4 Interface Conditions

To study whether the modifications of the prosthetic interface modified the phantom sensations, the participant was asked to walk for 2 min on a treadmill for each interface condition and then to describe in detail his phantom sensations. A treadmill protocol was preferred to an over-ground walking protocol to control the speed and optimize the regularity of the walking cycles. Preferred speed was determined on the ground during a timed walk over a distance of 10 m while wearing the semi-rigid socket without the patch. For safety reasons, the treadmill walking speed was set at 80% of the preferred speed, and the participant had to hold on lightly to the bars of the treadmill. The same speed (i.e., 3 km/h) was used for all conditions. The conditions were, in order: rigid socket without a patch, rigid

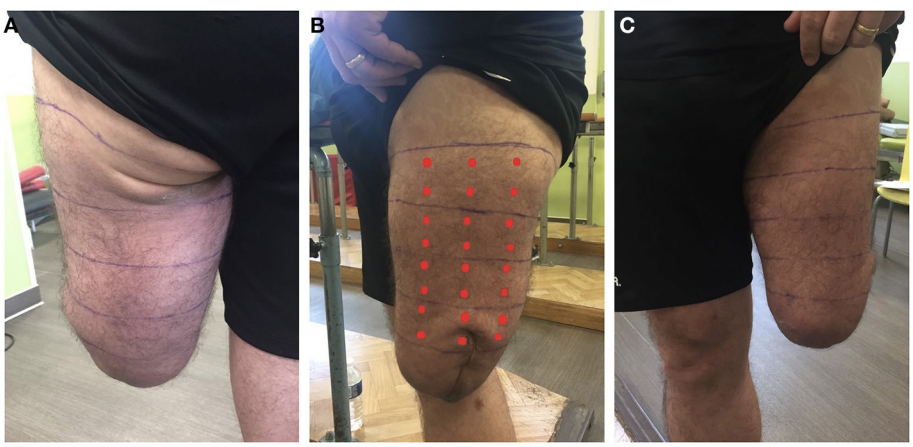

FIGURE 2 | Posterior **(A)**, lateral **(B)**, and anterior **(C)** views of the residual limb showing the 4 horizontal strips drawn for mapping. The most proximal horizontal delimitation was placed just under the groin and drawn perpendicular to the vertical axis of the residual limb. In each strip, 24 points around the leg were tested. The red dots in B represent the distribution of the stimulated points on one view. The part of the residual limb below the most distal strip was not tested as it was numb.

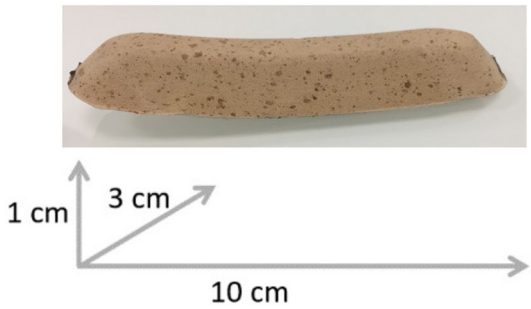

FIGURE 3 | The cork patch and its dimensions used to locally modify the prosthetic interface. It was 10 cm long, 3 cm wide, and 1 cm thick, rounded on the edges that touched the skin to avoid high pressure on the skin at these edges.

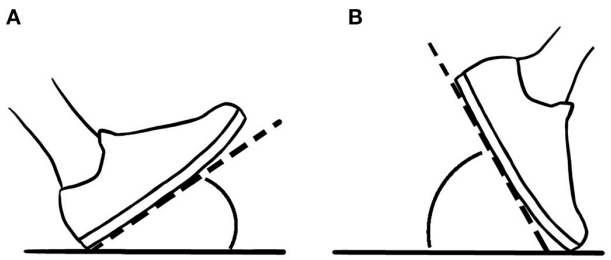

FIGURE 4 | Illustration of **(A)** heel strike angle and **(B)** toe-off angle. The heel strike angle is the angle between the ground and the foot when the heel landed on the ground at the beginning of the stance phase. The toe-off angle is the angle between the foot and the ground when the toes come off at the end of the stance phase.

socket with a patch, semi-rigid socket without a patch, and semi-rigid socket with a patch. A break of 10 min was taken between conditions. To study whether spatiotemporal parameters were modified by the different conditions, two Physilog® (Gait Up®) inertial units were placed on the lateral sides of the participant's shoes. The recording frequency was 128 Hz.

Variables

The phantom sensations were described qualitatively by the participant according to their nature and their location. Different gait parameters were analyzed for both the prosthetic and the contralateral limb: heel strike (**Figure 4A**) and toe-off (**Figure 4B**) angles, and duration of the double support phase. The latter corresponded to the percentage of the total duration of a gait cycle that both feet were on the ground. The first 3 gait cycles as well as the last minute of recording were removed from the data analysis, so a total of 38 gait cycles were analyzed for each interface condition.

Study Size

In this study, the main objective was to show that it is possible to modify the phantom sensations of an amputee by intervening on the socket, either locally or globally. This is an exploratory study to investigate the feasibility of this method of modifying phantom sensations with a single participant, before developing a larger scale study based on this method. Therefore, only one participant was included.

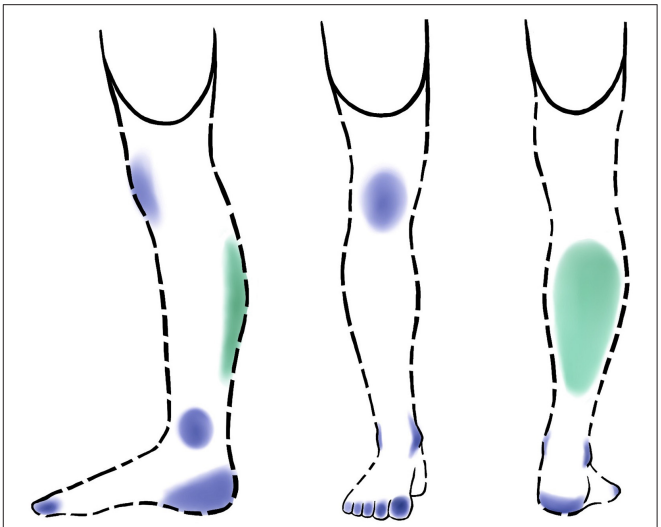

FIGURE 5 | Location of the participant's phantom sensations and the nature of the sensations (green: muscle contractions; blue: tingling) in a sitting position and without a prosthesis. In the standing position with a prosthesis and when walking, the areas are perceived more intensely and the locations are similar, except for the knee which is no longer perceived. Solid outline: residual limb; dashed outline: amputated limb.

Statistical Methods

R studio software (Version 1.3.1) was used to perform statistical testing on the gait parameters. As the data were from the same participant, a permutation ANOVA (27) was performed with 2 factors (type of socket and presence of patch).

RESULTS

Descriptive Data

Phase 1: Exploration of Phantom Limb Sensations With a Semi-structured Interview

The participant had permanent non-painful phantom sensations since his amputation. In a **seated position** without his prosthesis, he described permanent phantom sensations in the foot with tingling in the toes (especially the hallux), the heel, and the ankle (**Figure 5**). The knee was perceived occasionally and not very intensely. The participant perceived global "muscular contractions" in his calf. He did not perceive the anterior part of the leg. When **seated wearing his usual prosthesis** with the semi-rigid socket, besides the knee that disappeared, these phantom sensations remained permanent but intensified, especially for the toes. When **standing and walking**, this intensity again increased. Overall, these results show that the phantom sensations were influenced by wearing the prosthesis and the pressure applied on the residual limb.

At the end of the interview, the eventual presence of RS was explored by palpation of the residual limb. When an area on the

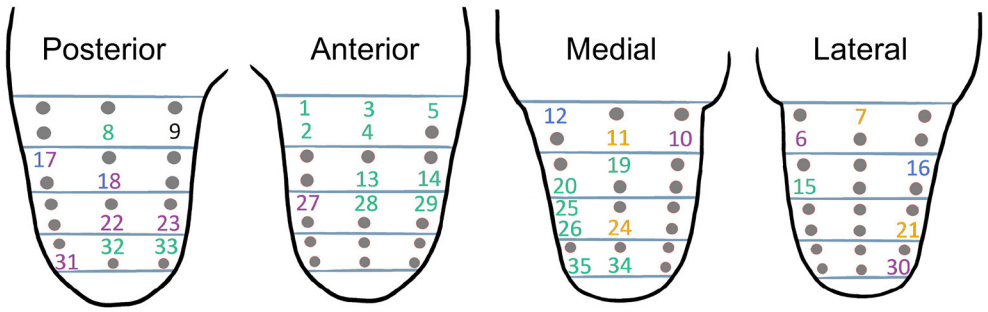

FIGURE 6 | Mapping of referred sensations (RS) on 4 sides of the residual limb (from left to right: posterior, anterior, medial and lateral side). The grey dots indicate the areas that did not give rise to the RS; the numbers those that did. The color of the numbers indicates the nature and the location of the RS (green: contraction of the calf; yellow: influx in the calf; blue: tingling in the foot; purple: influxes in the foot). Pressure on area 9 evoked a sensation that was too brief and weak to be identified, and on areas 17 and 18 an impulse and tingling in the foot.

posterior side of the residual limb was pressed, the participant felt his phantom calf more intensely. Another area was found on the medial side on which pressure induced perception of his phantom foot arch, whereas he had not perceived it spontaneously. A more detailed exploration was therefore carried out in Phase II to map his RS more methodically.

Phase 2: Mapping of Referred Sensations

Phantom sensations were modified by pressure for 35 of the 96 stimulated areas, 9 were localized posterior on the residual limb, 10 anterior, 10 medial, and 6 lateral (**Figure 6**). The RS were more intense and precise localized muscle contractions in the calf and more intense tingling in the hallux, sensations on phantom segments that the participant already perceived. In addition, he perceived "influxes" in the calf and the foot, as well as tingling in the arch, dorsum, and medial side of the foot, the latter being sensations and phantom segments that the participant had not perceived spontaneously.

FIGURE 7 | The cork patch placed on the posterior side of the rigid **(A)** and semi-rigid socket **(B)** on stimulation areas 8 and 18 (see **Figure 6**).

Main Results

Phase 3: Global and Local Modifications of the Prosthetic Interface

The patch was placed in both the rigid and semi-rigid sockets on areas 8 and 18 of the map (posteriorly and proximally on the residual limb, **Figure 7**) for which pressure induced modification of sensations of calf muscle contractions, and foot influxes and tingling in the arch of the foot (see **Figure 6**).

Phase 4: Description of the Phantom Limb Sensations When Walking and Recording of Gait Parameters for Each of the 4 Interface Conditions

Concerning the phantom sensations, for the foot, the type of interface changed the vividness and location of the tingling in the foot. For both sockets, wearing the patch, the tingling in the foot was more vivid and distributed throughout the foot rather than being concentrated or more intense in the forefoot (**Table 1**). With the semi-rigid socket with a patch, the participant also perceived the arch of the foot, whereas he did not perceive it in the other three conditions. The localization and intensity of the RS were stable. Concerning the calf contractions, wearing the patch, the participant perceived his calf much more vividly, both through an increase in the felt contraction intensity and the distinction of the parts of the calf. Moreover, for both sockets, the patch induced sensations of calf contractions that varied with the gait cycle. Indeed, when the prosthetic heel hit the ground, he felt a contraction at the distal part of the phantom calf that moved upward toward the end of the stance phase. This gave him a sensation in the phantom calf that varied simultaneously with the progress of the stance phase. So, the sensations in the calf were dynamic, which was not the case for the calf without a patch and for the foot. When asked to rank the 4 conditions, the participant had a clear preference for the semi-rigid socket with patch and even asked to keep the patch at the end of the experiment.

Concerning the gait analysis, the type of socket and the presence of the patch influenced the **heel strike angle** similarly for both the prosthetic and the contralateral limb (**Figure 8**). Whether the patch was present or not, this angle was found to be larger for the semi-rigid socket than for the rigid one ($p < 0.001$). But the effect of the patch was different depending on the

TABLE 1 | Description of the participant's phantom sensations according to the 4 prosthetic interface conditions.

	Rigid		Semi-rigid	
	Without patch	**With patch**	**Without patch**	**With patch**
Foot tingling	Restricted to the hallux Constant	Diffused over the toes and the heel Constant	Diffused over the foot, forefoot most intense Constant	Diffused over the whole foot Constant
Calf contractions	Global Constant	Focal Dynamic	Global Constant	Focal Dynamic
Ranking	4	3	2	1

The detailed elements are the location and evolution during gait cycles of foot tingling and calf muscle contractions. "Constant": Non-varying intensity and localization during gait cycles. "Dynamic": Intensity and localization consistently varying during gait cycles. The last row contains the participant's ranking of the interface conditions, with 1 being the preferred.

type of socket ($p < 0.001$). Indeed, the patch in the rigid socked decreased the heel strike angle, whereas in the semi-rigid socket it increased this angle (prosthetic limb: $p < 0.05$; contralateral limb: $p < 0.001$). Note that the condition preferred by the participant (semi-rigid socket with patch) had the largest heel strike angle for both limbs.

The interface condition influenced the **toe-off angle** differently for the prosthetic and the contralateral limb. For the prosthetic limb, regardless of the presence of the patch, the semi-rigid socket induced a larger angle than the rigid one ($p < 0.05$). For both types of sockets, the presence of the patch increased the angle ($p < 0.001$). So, the toe-off angle for the prosthetic limb was largest for the participant's preferred condition. For the contralateral limb, only the type of socket influenced the toe-off angle that was smaller when wearing the semi-rigid socket ($p < 0.001$).

Finally, both the type of socket and the presence of the patch influenced the **duration of the double support** phase. For both types of sockets, the presence of the patch diminished the duration ($p < 0.05$). Regardless of the presence of the patch, the semi-rigid socket induced a longer duration ($p < 0.001$).

DISCUSSION

Key Results

This single case preliminary study showed that phantom sensations during walking can be modified by interventions on the socket interface. The participant reported that the nature of the phantom sensations and the concerned segments varied among the prosthetic interface conditions. These modifications were accompanied by changes in some spatiotemporal parameters.

Limitations

Rehabilitation professionals and a few rare articles studying the link between the prosthetic interface and walking, acknowledge that mechanical stresses related to the socket influence walking (28) without even considering the phantom limb. Thus, although it seems likely, given the participant feedback, we cannot at this time assert that the changes in gait parameters are causally related to RS. The inclusion of a group of amputee participants without phantom sensations, for whom the same mechanical modifications on the interface of the prosthesis will be performed, will allow us to determine whether the phantom sensations modified by the interface did indeed influence the gait parameters or whether the modifications to the interface alone are responsible for this phenomenon.

Even if the modifications of the interface prove to be the cause of the modifications in the gait parameters, this would not affirm an increase in the quality of his gait. As the aim of the study was not to understand the walking strategy, we only recorded a limited number of spatiotemporal parameters. This does not allow us to assert that the entire gait strategy was modified by the intervention on the socket and phantom sensations, nor to explain the observed changes in certain spatiotemporal parameters, let alone their possible consequences in the long term. But even if we had recorded more parameters, the current state of knowledge on prosthetic locomotion does not allow us to define the optimal gait in lower limb amputees. This is why further analysis of gait but also of more cognitive variables such as mental workload (29), will allow us to determine whether (dynamic) phantom sensations can be used as somatosensory feedback and be useful for walking.

Interpretation

The sensations induced by the patch were not simply the sum of the sensations caused by the individual pressures on areas 8 and 18 of the map. Indeed, the patch induced a more complete phantom foot, and the calf sensations became dynamic. This could be related to the fact that, first, the mapping was performed by delivering a focal pressure on the residual limb, whereas a socket induces a global pressure on the whole residual limb. Second, the mapping was performed at rest, whereas the effect of the interventions on the RS concerned walking. This induced highly dynamic variations in the pressure distribution in the socket, and thus cyclically altered the amount of additional pressure applied by the patch on the residual limb. The participant in this study already had many permanent phantom sensations during walking, which he reported using daily. However, we found an intervention that created *dynamic* phantom sensations in the calf varying systematically with the phases of the gait cycle. These dynamic sensations were particularly positive for the present participant who evaluated the conditions with the patch on top.

Evaluations of spontaneous phantom limb sensations (nature and localization) and the stimulation type and areas on the residual limb inducing RS are necessary to propose adapted modifications of the socket interface. For the participant of the present study, the stimulus inducing RS appeared to be pressure, but it is known that RS can be induced

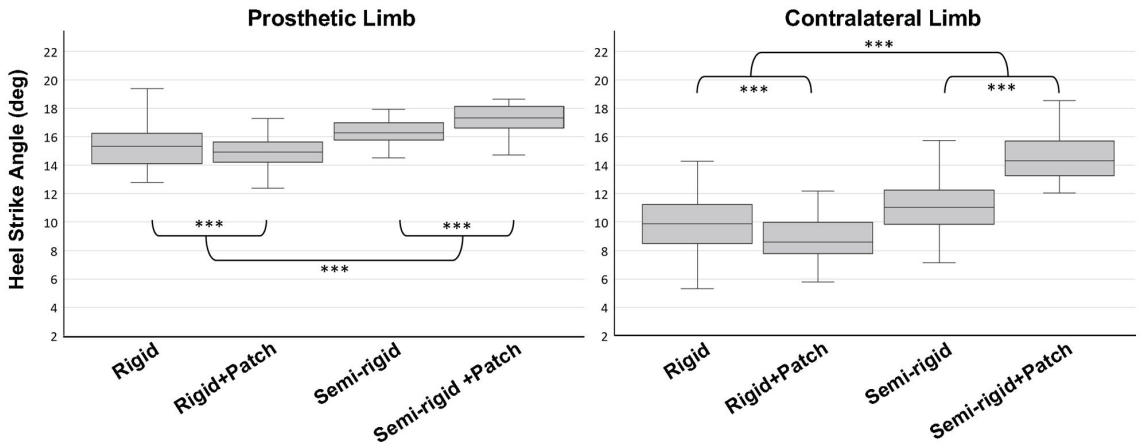

FIGURE 8 | Heel strike angle in the 4 prosthetic interface conditions for the prosthetic (left) and the contralateral limb (right). For both limbs, the effect of the patch on the heel strike angle depended on the socket type ($p < 0.001$). ***$p < 0.001$.

by other-than-static-pressure types of stimuli such as light touch (23), vibration (25), or electrical stimulation (20, 25). These stimuli have already been used to give feedback about phases of the gait cycles (11, 30). Yet, in these substitutive solutions, the stimulations were not located on the referred sensation map and thus gave only rise to a perception of the stimulus and not of the limb in motion. This means that the association between the nature of the stimulus and what it is supposed to represent must be learned. This is not the case in our approach that allows restoration of the perception of the limb evolving in action through RS. Our new approach could use similar gait phase detection and stimulation techniques as in substitutive solutions, but if the stimuli are delivered respecting the referred sensation map, they will induce the perception of the limb in action without needing a learning phase.

The modifications on the interface not only had a clear impact on the participant's phantom sensations but were also accompanied by a change in gait parameters. Indeed, the presence of the patch and the type of socket had a complex effect on several spatiotemporal gait parameters with (i) the influence of the patch depending on the type of socket, and (ii) the influence differing between the contralateral and prosthetic limb. Interestingly, the participant naturally expressed a strong preference for the semi-rigid socket with patch condition, which was the condition found with the most extreme values of the gait parameters among all conditions. Yet, the relationship between RS and the change in gait parameters is not clear. It is questionable whether the participant changed his gait parameters because the perception of his phantom leg in action allowed him to do so, or, on the contrary, because he liked the sensations during walking and the change in gait parameters allowed him to have them. The increase in heel strike and toe-off angles could be in favor of the second hypothesis, as this could have resulted in greater pressure on the proximal posterior part of the socket, which was the area where the patch was positioned.

Overall, this study suggests that after ower limb amputation, it is possible to restore the perception of the limb evolving in action through RS by modifying the design of the socket. Currently, the form of the socket is only considered in relation to support constraints but this new approach may lead to rethinking the personalization of the interface in terms of its relationship with the phantom limb. We encourage therefore to consider phantom sensations from the beginning of rehabilitation.

Generalizability

As this preliminary study was based on a single case, our interpretations and conclusions cannot yet be generalized to the entire population of lower limb amputees. However, this feasibility study of the method of modifying phantom sensations perceived during walking is encouraging and now allows us to continue this research work with a larger population by improving the methodology taking into account the identified limitations.

DATA AVAILABILITY STATEMENT

The raw data supporting the conclusions of this article will be made available by the authors, without undue reservation.

ETHICS STATEMENT

Ethical approval was not provided for this study on human participants because this was a pilot experiment to determine whether an in-depth study should be set up. This is done in total agreement with the participant and under supervision of the medical staff of the rehabilitation clinic following the patient. The orthoprosthetist of the patient did the interventions on the prosthesis. Given the results that we obtained, the ethical approval is in preparation in order to have a ethical review for the future study involving more patients. The participant provided his written informed consent to participate in this study.

AUTHOR CONTRIBUTIONS

LB performed all experiments and wrote the first version of the paper. SL conceived the prostheses sockets and together with CM helped during the experiments. SM helped analyzing the gait parameters. AT, IL, and JP contributed to theoretical considerations. JD is the senior researcher leading the project and finished the article together with LB. All authors contributed to the article and approved the submitted version.

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Temporary Botulinum Immobilization of Residuum Muscles for Facilitation of the Initial Ingrowth of Skin to the Porous *Skin and Bone Integrated Pylon* in the Technology of Direct Skeletal Attachment: Large Animal Model

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Enhancing the technology of bone-anchored limb prosthetics, we present a modified porcine model for developing an infection-free integration between the skin and a percutaneous bone implant. The deeply porous Skin and Bone Integrated Pylon (SBIP) presented an infection-free skin-implant interface both after implantation into the dorsum and after implantation into the residuum after below-knee amputation. However, deep ingrowth of skin into the porous cladding of the SBIP was achieved better in the dorsal procedure, while implantation to the residuum sometimes developed a stoma, probably due to the high mobility of the skin and soft tissues in the pig's thigh. Uncontrolled high skin mobility during the first week after implantation constituted a limitation for the porcine animal model, which we tried to address in the current study. As our previous studies showed that casting of the leg residuum did not sufficiently limit the skin's movement around the implant, we tested a modified protocol of the implantation, which included injection of botulinum toxin into the thigh muscles. During the course of the study, we identified proper botulinum toxin componentry, dosage, and the period after injections to achieve a maximal effect of immobilization of the muscles affecting skin movements. To verify the immobilization, we used kinetic data on the asymmetry of loading during gait with the Strideway System, Tekscan, Inc., Boston, MA, USA. We found that injections in the four muscles of the distal thigh of the left hind leg with MYOBLOC® (rimabotulinumtoxinB; 5,000 units/muscle) were sufficient to provide noticeable immobilization by the fourth week after the procedure. This conclusion was made based on the analysis of the dynamics of asymmetry in vertical ground reactions on the injected (left hind) and uninvolved (right hind) legs during gait over an instrumented walkway.

Keywords: direct skeletal attachment, porcine model, skin immobilization, botulinum injections, osseointegration, body-implant interface

INTRODUCTION

Bone-anchored limb prostheses offer a number of advantages over socket-based prostheses (1). The technology of osseointegration relies on the integration of the residuum's bone with the titanium implant and traces its origins to the 1950s in Sweden by Dr. Per-Ingvar Brånemark (2, 3).

A problem with this technology is the still high infection rate at the interface of the skin with the implanted fixture (4–6).

Percutaneous porous devices used in bone-anchored prostheses have the potential for initial integration with the skin, as demonstrated in animal studies by various research groups (7–9). Our studies have also investigated porous implants for direct skeletal attachment, focusing on the ability of implants to invite and sustain deep skin and bone ingrowth to promote an infection-free body-device interface while maintaining the required mechanical strength. The implant we developed with such features is called the Skin and Bone Integrated Pylon (SBIP) (8, 10–13). The innovation of the SBIP lies in its patented combination of four key technological characteristics: *porosity*, *pore size*, *porosity volume fraction* (VF), and *particle size,* and a provision for the *passage for the wired neural interface,* and *protective silver coating* (10, 14).

The parameter most distinct from the prior art, which most meaningfully distinguishes the SBIP from other systems, is the *porosity VF*, which quantifies how porous the implant is (formally, *VF* is the ratio of the volume of the porous portion to the entire volume of the device).

As the SBIP implants have been designed to encourage and enable deep skin permeation, there is a critical and vulnerable period—between implantation and full permeation—that requires methodological advances. Until the surrounding skin cells remodel within all of the implant's pores, special care to minimize the skin movements around the implant is required to protect the still non-occupied pores from bacterial infiltration (15–17). Minimizing skin movements during the initial period after transdermal implantation is especially important in the studies with large animals (pigs), since the activity of the massive musculature in the residuum and above may mechanically pull out the skin around the implant.

Our previous studies with pigs (18) showed that deep and sustainable ingrowth of skin into the porous cladding of the SBIP can be achieved after implantation into the pig's dorsum. As to implantation into the residuum of the leg, the skin developed a stoma around the implant (15, 17). There is an excess of the movable skin and soft tissues in the pig thigh; simple casting did not successfully immobilize the skin while the skin seal was developing.

Since our overall intention is to establish a sustainable and safe skin seal to provide natural barriers against infection, we tested here a modified implantation protocol. The modification is the inclusion of pre-implantation injections of botulinum toxin to temporarily immobilize the muscles that affect the movement of skin in the implantation zone.

Botulinum toxins are approved by the Food and Drug Administration (FDA) for application in human patients (19) and are frequently used in patients with spasticity of the upper and lower limbs due to upper motor neuron disorders, spinal cord injuries, multiple sclerosis, strokes, brain injuries, and cerebral palsy (20, 21). Botulinum toxin inhibits the release of acetylcholine at the neuromuscular junction, reducing the contraction of the muscle (22).

Fewer reports are extant on botulinum toxin applications in pigs (23). That makes it necessary to judiciously select the type of toxin and its dosage, which may differ from those recommended for humans (24–26).

In the current pilot study with three animals, we calibrated both the dose and optimal timing for the implantation, which is when the botulinum toxin reaches its maximum effect. This paper presents the leading hypothesis, study design, and outcomes of the study.

STUDY DESIGN

The study protocol #DB-633, "Effect of botulinum neurotoxin serotypes A or B injections into thigh musculature of a swine," was approved by the Institutional Animal Care and Use Committee (IACUC) at DaVinci Biomedical Research Products, Inc., Lancaster, MA, USA and by the US Army Medical Research & Development Command (UASMRDC) Animal Care and Use Review Office (ACURO) on July 16, 2020, with further approval of Amendment 1 on March 25, 2021.

The purpose of this botulinum toxin study was to determine the period when the injection's immobilization effect was the greatest on the pig leg muscles. The contraction of these muscles can compromise the initial remodeling of the skin while a sustainable seal is developing after implantation of the transdermal implant into the leg's residuum.

The best timing for the implantation is when the immobilization effect is strongest. The asymmetry of loading between the uninvolved hind leg and the hind leg with injected botulinum toxin can be used to detect the maximum immobilization.

The intensity of immobilization was quantified by an Asymmetry Index (AI) calculated from quadruped gait analysis data obtained with the Strideway System, Tekscan, Inc. Boston, MA, USA. A standard set of data is illustrated in **Figure 1**. For each of the gait trials, the Strideway software, among other parameters of gait, generates a Symmetry Table as the ratios of the magnitudes of the various parameters for left and right legs. **Figure 1** depicts a Symmetry Table associated with one of the five gait trials (B05) conducted with animal #3 in 4 weeks after botulinum toxin injection. The ratio "Max Force Left Hind/Rights Hind" (encircled in red square) is a parameter we called AI. We have selected this parameter for characterization of the inhibiting effect of botulinum injection on the activity of the leg muscles. An ideal magnitude of AI in sound gait, when the load on the right and left legs is equal, is 1.00.

Reports in human applications of botulinum toxin injections indicate that the mean time to peak effect is ranging from 2 weeks to 3.7 (SD ¼ 2.4) weeks, and that treatment effects declined at a mean of 9.3 (SD ¼ 4.0) weeks (27, 28).

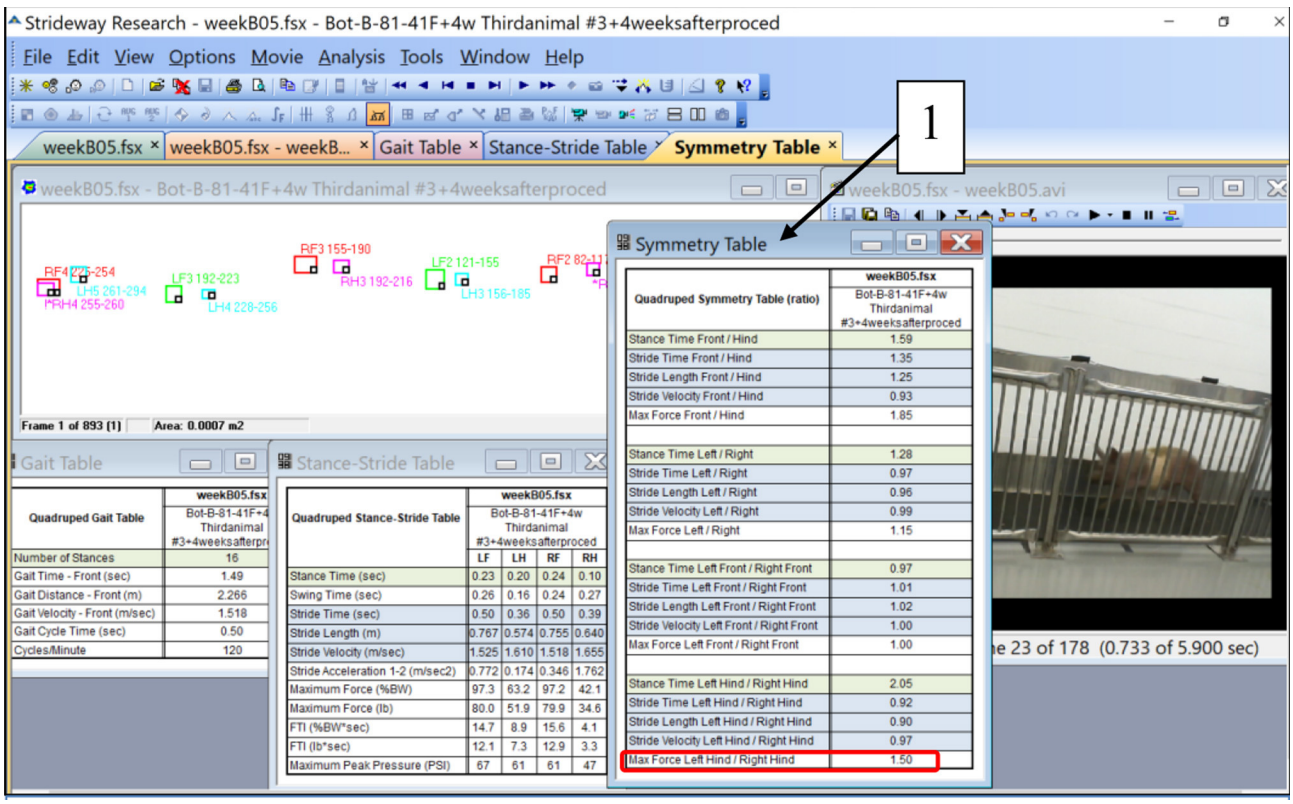

FIGURE 1 | Illustration of the parameters of quadruped gait calculated by the Strideway software including a Symmetry Table (1). A ratio "Max Force Left Hind/Right Hind" at the bottom of the Symmetry Table (encircled with red) was taken as a parameter called "Asymmetry Index (AI)" (see **Table 2** and the graph in **Figure 3**).

We hypothesized that within this interval, neuro control over the muscle-coordinated activity during the gait cycle will change the magnitudes of the loading of the injected leg, as detected in the increase in the IA.

Thus, the purpose of this botulinum toxin study was to confirm this hypothesis or to make the necessary modifications in the type of botulinum toxin or its dosage.

METHODS

Procedures

We injected botulinum toxin A (Xeomin®), Merz Pharma GmbH & Co., Dessau, Germany, an Incobotulinum product, equivalent to Botox® and Dysport® (29), and compared its effect with Botulinum toxin type B (MYOBLOC Elan Pharmaceuticals, Inc., San Francisco, CA, USA), which showed better desired effect in pigs compared to toxin A in pig masseter muscles (23). The injections were dosed at 8 units/kg, similar to human pediatrics and equivalent to the maximum allowed dose by the FDA in children to the lower limb (19).

Injections were performed using ultrasound guidance into the rectus femoris, vastus lateralis, vastus intermedius, and vastus medialis of the pigs in order to increase adherence of the below-knee prosthesis (see **Table 1**).

Application of the botulinum toxin treatment included injections into the distal musculature of the hind limb of the pig; daily monitoring during first 2 weeks and weekly monitoring of behavior and locomotor activity of the animal; gait analysis of the pre-procedure and following 2, 4, 6, 8, 10, and 12 weeks after the injection.

TABLE 1 | Injected muscles, toxin type and dosage.

Injected muscles	Toxin type and dosage		
	IncobotulinumA (Xeominò)	RimabotulinumtoxinB (Myobloc®)	
	Animal 1 No. 1090	*Animal 12 No. 1143F*	*Animal 3 No. 81-141F*
Rectus femoris	2.0 mL (100 units)	4.5 mL (7,500 units)	2.5 mL (5,000 units)
Vastus lateralis	2.0 mL (100 units)	4.5 mL (7,500 units)	2.5 mL (5,000 units)
Vastus intermedius	4.0 mL (200 units)	4.5 mL (7,500 units)	2.5 mL (5,000 units)
Vastus medialis	2.0 mL (100 units)	4.5 mL (7,500 units)	2.5 mL (5,000 units)
Gluteus maximus	2.0 mL (100 units)	N/A	
Total units injected	12.0 mL (600 units)	18 mL (30,000 units)	10 mL (20,000 units)

Figures 2A–F illustrate the procedure of the Botulinum study in Animal 1 No. 1090. Xeomin® (**Figure 2A**),

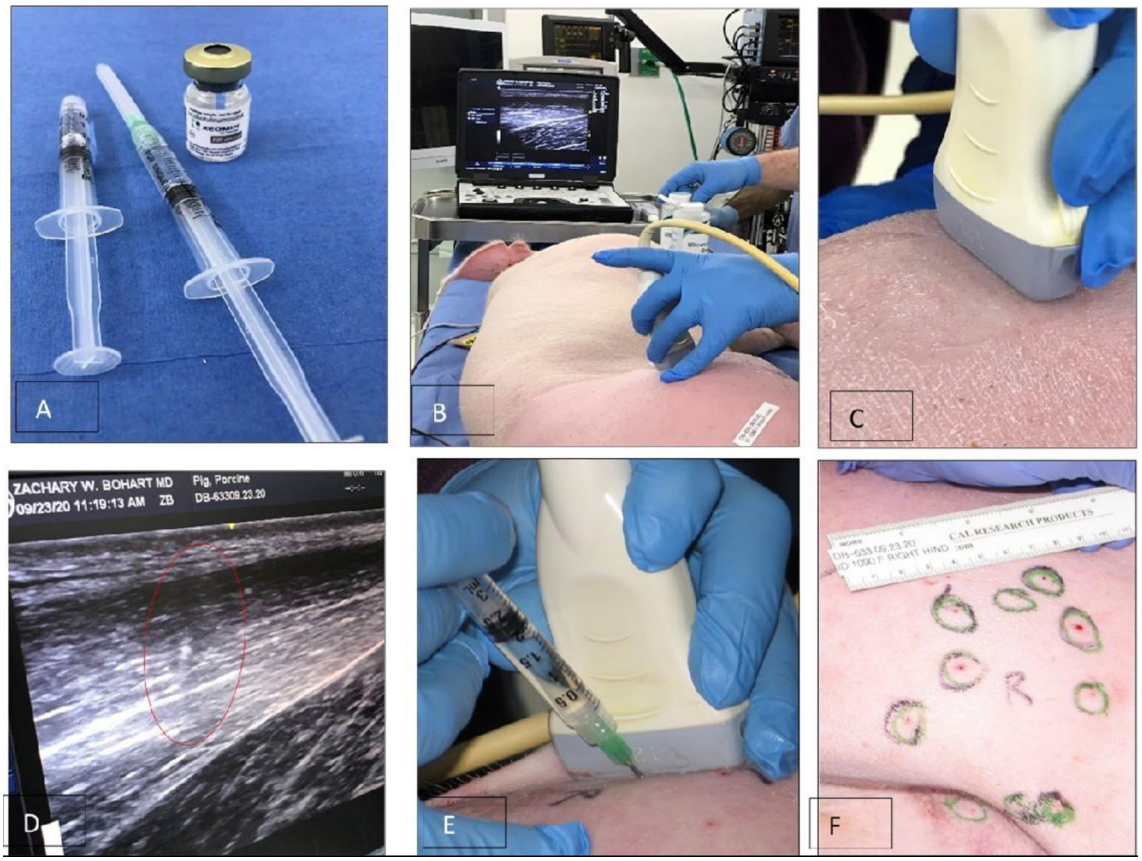

FIGURE 2 | A study on immobilizing skin and muscles before the osseointegration procedure for a better integration of tissues at the skin-implant (SBIP) interface. **(A)** Xeomin®, an incobotulinum toxin A product equivalent to Botox® and Dysport®. **(B)** GE Ultrasound laptop machine for guidance of injections. **(C,E)** Finding a spot for injection with visual confirmation **(D)**. **(F)** Schematics of the injection spots in the study with the Animal 1-No. 1090, 9-23-20.

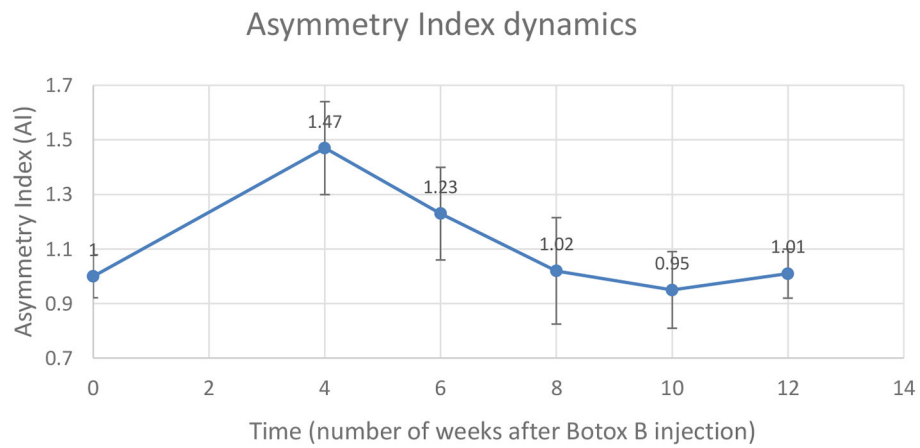

FIGURE 3 | Changes in the Asymmetry Index (AI) as a ratio of maximal vertical ground reaction on the injected left leg (LH) to the uninvolved right leg (RH).

GE Ultrasound laptop machine for guidance of injections (**Figure 2B**). Finding a spot for injection by moving the transducer with visual confirmation on the screen of the GE Ultrasound machine (**Figures 2C–E**). Schematics of the injection spots in the study, 9-23-20 (**Figure 2F**).

Outcomes

Animal 1 No. 1090

Application of incobotulinumtoxinA was performed between September and December of 2020 by injection of botulinum toxin A to the distal thigh musculature of the right leg.

Outcomes

The Xeomin® was ineffective. There was no muscle weakening and therefore, no effect on animal gait. For the duration of 3 months post-injection and 3 months wash-out period, the animal was normal. The animal's gait was normal throughout, with symmetrical kinematic and kinetic data compared between the involved and uninvolved legs.

The conclusion was made based on results of the consecutive gait analysis that Botulinum toxin A does not provide sufficient immobilization of the leg muscles and that a new injection with Botulinum toxin B was suggested with a modified dosage.

Animal 2 No. 1143F

Since the previous injections of incobotulinumtoxinA proved ineffective for blocking muscular contraction, the second cycle of application of botulinum toxin B, a different serotype of botulinum toxin treatment [MYOBLOC® (rimabotulinumtoxinB)], was performed on January 23, 2021.

Ultrasound-guided muscular injections were entered into the distal thigh of the right hind leg, with MYOBLOC® (rimabotulinumtoxinB; 5,000 units/1 ml) diluted from 1 to 3 ml using injectable saline. Four (4) muscles were each injected with 7,500 units/muscle.

The animal recovered well from the injection procedure. The animal was observed 2x daily. Observations of animal and injection sites were normal. On day 4, during AM checks, the limb appeared normal. The animal was found lying down and not eating. The animal was unable to stand and was non-responsive. After a consult with Attending Veterinarian, it was determined that there was toxicity. Animal was referred for unscheduled euthanasia. The animal was euthanized the same day.

Necropsy Notes

Temperature: 101.9F; heart rate: 120; respiratory rate: 20; capillary refill time: >4 s. Animal was unable to stand, lethargic. Injection sites were normal. The animal was found laterally recumbent, was paretic in the hind and front end and was slightly cyanotic.

Animal 3 No. 81-141F

The third animal received a smaller dosage of rimabotulinumtoxinB than Animal 2, recovered, and was tested with the Strideway gait analysis system.

Application of Botulinum treatment was performed on May 19, 2021.

UV-guided muscular injections were performed to the distal thigh of the left hind leg with MYOBLOC® (rimabotulinumtoxinB; 5,000 units/1 ml) diluted from 1 to 0.5 ml using injectable saline. Four (4) muscles were each injected with 5,000 units/muscle.

TABLE 2 | Asymmetry Index dynamics over time after injection.

Animal #3		
Time (weeks) after injection	**Asymmetry index (AI)**	
	Mean	**STDEV**
0	1.00	0.08
4	1.43	0.21
6	1.23	0.58
8	1.02	0.19
10	1.03	0.28
12	1.01	0.09

Animal No. 81-141F had an uneventful recovery.

Animal was observed 2x daily. The injection sites were normal throughout the survival period. On day 6, post- injections the animal started to become paretic. This paresis lasted 6 days during which the animal was tube fed and intermittently placed in a Panepinto sling. The animal made a full recovery and was able to complete all the gait analyses.

Weekly monitoring of behavior and locomotor activity of the animal demonstrated recovery from the injection and return to regular ambulation with the greatest asymmetry in kinematic and kinetic data at week 4 after injection procedure.

Gait analysis was performed six times: 2 days pre-procedure as a baseline, and 4, 6, 8, 10, and 12 weeks after the injection procedure. The dynamics of the AI is shown in **Table 2** and is illustrated in the chart (**Figure 3**). A distinct increase of AI occurred at week 4 after the injection. By weeks 8–12, the AI was recovered to the initial symmetry in loading of both hind legs.

A mean number of stance cycles was 15 (SD 1) and a mean of the walking distance was 2.17 (SD 0.10) m. The maximal AI at week 4 indicated that the loading on the injected leg by that time exceeded the loading on the uninvolved leg by 47 ± 17%. Within this interval, the neurocontrol over the muscle-coordinated activity during gait cycle was affected by the Botulinum toxin, which did not allow the leg to be lifted as quickly as the contralateral leg, which resulted in the higher magnitude of normal ground reaction.

DISCUSSION

We anticipated that by immobilizing the distal thigh muscles ∼4 weeks before the transdermal implantation, the initial ingrowth of skin into the porous cladding will progress without being torn off by muscular movement. By that, more favorable conditions are anticipated for the creation of the skin seal at the implant-skin interface as a natural barrier against infection.

We did not consider differences among tested animals (e.g., in terms of the body morphology and sex) due to their small number, which constitutes a limitation of this pilot study.

We will investigate the benefits of the pre-implantation Botulinum injections in our further studies in bone-anchored prosthetics with this modified porcine model. The model with pre-implantation botulinum toxin injections may have higher translational value than the regular one, considering existing FDA-approved Botulinum applications in humans.

CONCLUSIONS

1. Injections with incobotulinumtoxin^{A} (Xeomin®) were ineffective at inducing any form of muscle weakness with effect to gait.
2. MYOBLOC® (rimabotulinumtoxinB) injections proved toxic with the first dosage applied. A range finding study was recommended to identify the optimal dose to induce muscle weakness.
3. A smaller dosage of MYOBLOC® (rimabotulinumtoxinB) showed safe outcomes of the injection and demonstrated the effect expected—asymmetry (47 ± 17%) in loading between affected and non-affected limbs 4 weeks after the injection (**Figure 2**) compared to baseline recording (**Figure 1**). Further observations showed recovery of the symmetry in gait parameters: as 23 ± 21% in 6 weeks, 2 ± 21% in 8 weeks, and 3 ± 21% 10 weeks after the injection procedure (see **Table 2**, **Figure 2**).
4. Limitations of the study include a small number of animals and the pilot selection of the dosage is found effective. For addressing these limitations further studies are suggested.

DATA AVAILABILITY STATEMENT

The original contributions presented in the study are included in the article/supplementary material, further inquiries can be directed to the corresponding author.

ETHICS STATEMENT

The animal study was reviewed and approved by IACUC DaVinci Biomedical research Products and by ACURO UASMRDC.

AUTHOR CONTRIBUTIONS

ZB participated in the study design and injection methodology. CC participated in injection procedures and the translational value of the modified protocol. DM, RV, MV, and LC provided animal husbandry, enrichment, and veterinary care. MP contributed to the study design and gait analysis. All authors contributed to the article and approved the submitted version.

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Limb Prostheses: Industry 1.0 to 4.0: Perspectives on Technological Advances in Prosthetic Care

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Technological advances from Industry 1.0 to 4.0, have exercised an increasing influence on prosthetic technology and practices. This paper explores the historical development of the sector within the greater context of industrial revolution. Over the course of the first and up the midpoint of the second industrial revolutions, Industry 1.0 and 2.0, the production and provision of prosthetic devices was an *ad hoc* process performed by a range of craftspeople. Historical events and technological innovation in the mid-part of Industry 2.0 created an inflection point resulting in the emergence of prosthetists who concentrated solely on hand crafting and fitting artificial limbs as a professional specialty. The third industrial revolution, Industry 3.0, began transforming prosthetic devices themselves. Static or body powered devices began to incorporate digital technology and myoelectric control options and hand carved wood sockets transitioned to laminated designs. Industry 4.0 continued digital advancements and augmenting them with data bases which to which machine learning (M/L) could be applied. This made it possible to use modeling software to better design various elements of prosthetic componentry in conjunction with new materials, additive manufacturing processes and mass customization capabilities. Digitization also began supporting clinical practices, allowing the development of clinical evaluation tools which were becoming a necessity as those paying for devices began requiring objective evidence that the prosthetic technology being paid for was clinically and functionally appropriate and cost effective. Two additional disruptive dynamics emerged. The first was the use of social media tools, allowing amputees to connect directly with engineers and tech developers and become participants in the prosthetic design process. The second was innovation in medical treatments, from diabetes treatments having the potential to reduce the number of lower limb amputations to Osseointegration techniques, which allow for the direct attachment of a prosthesis to a bone anchored implant. Both have the potential to impact prosthetic clinical and business models. Questions remains as to how current prosthetic clinical practitioners will respond and adapt as Industry 4.0 as it continues to shape the sector.

Keywords: prosthetics, technology, Industry 4.0, innovation, history

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INTRODUCTION

The term Industry 4.0 is frequently used enthusiastically to describe a new wave of exponential innovation that will revolutionize the world as we know it and, with it, the field of orthopedics, including prosthetic devices (1, 2). The term itself signals it is not the first, inviting the questions "What are the industrial revolutions?" and "What do they have to do with prosthetics?"

It has become common to describe technological evolution within a framework of industrial revolutions, which are used to denote eras of significant change in how goods are designed and produced or how technological developments change products and processes. Initially it was used to describe the transition from an agrarian society to an industrial one beginning in the mid 1700's. It is now used to describe four eras: The Industrial Revolution (Industry 1.0), The Technological Revolution (Industry 2.0), The Digital Revolution (Industry 3.0) and the Physical, Digital, and Biological Convergence (Industry 4.0) (2, 3).

This paper aims to present examples from the history of prosthetic technology evolution within the industrial revolution framework to highlight how current prosthetic provision clinical practice and business models could benefit from current technological innovations or could be significantly disrupted by it. It is hoped this paper will help current prosthetic service providers understand the need to be proactive in navigating these cross-sectoral changes as the engineers and technology developers driving them begin to insert themselves into the prosthetic provision process.

It should also be noted that the history presented focuses predominantly on developed nations, as the use of technology intensive prosthetic componentry is still concentrated in developed nations. However, Industry 4.0 is seen as having the potential to improve access to care globally. Initial improvements to access to care are already emerging in response to the Covid-pandemic (4). Specific to prosthetics, additive manufacturing to improve access to prosthetic technology in low- and middle-income countries is seen as an emerging area of research and development (5).

This paper traces the four industrial revolutions, identifying key themes and presenting some of the innovations that occurred in each era that eventually found their way into prosthetic design and practice. There are limitations to this approach, as each era does not have clearly defined start and end dates. And, because of the lag in adoption of new technologies and processes by the sector, along with non-liner technology development, it is perhaps better to think of progress in this sector as a spectrum with some overlap between eras (**Figure 1**).

It is in no way a complete history of either industrial revolutions or the prosthetics sector. Instead it seeks to link how the key themes of each era of industrial revolution eventually impacted, in some way, prosthetic design or practice. It also explores how technological change and historic events shaped the prosthetics sector actively, embracing innovation or change, as well as passively, where innovation or change occurred because it was no longer possible to maintain the past way of doing things.

[1]https://stokodesign.com/pages/how-it-works

INDUSTRY 1.0: INDUSTRIAL REVOLUTION

Industry 1.0, also known as the First Industrial Revolution or simply as the Industrial Revolution, began with the development of mechanisms to harness water and steam power to drive industrial machines in the eighteenth century. This period of industrialization shifted society's focus from agrarian to industrial giving rise to machinery that could produce goods that up to that point had been produced by hand. This was done in factories that, coupled with improved and more efficient transportation, allowed goods to be moved further and more cost effectively than before (3). Classic examples are water powered looms used to weave cloth in mills and steam trains that transported people and goods to factories and markets. Agriculture was also becoming mechanized, allowing more efficient food production and freeing farm workers to move to larger centers to work in mills and factories. This was not a revolution in the military sense. It was a revolution in the way people lived, worked and conducted business. At the start of Industry 1.0, the production of prosthetic limbs was craft based using locally available materials such as leather, wood and metal (6–9). Amputation was typically due to trauma which was often the result of warfare and few amputees could afford a device. Prosthetic limbs were not widespread and amputees improvised with what they had in order to ambulate. Prior to and including the period spanning Industry 1.0, little to no advancement in prosthetic design occurred over a period of centuries.

INDUSTRY 2.0: TECHNOLOGICAL REVOLUTION

Industry 1.0 began to transition to 2.0 at the mid to latter part of the nineteenth Century with some sources marking the First World War as the start of the Industry 2.0 (10). This period was marked by the invention of devices that could capture and store electrical energy leading to electrification of factories and an increasing use of mass production. At the same time many novel materials and inventions were created. Industry 2.0 is commonly referred to as the Technological Revolution.

Over the course of Industry 2.0 two wars left their mark on the prosthetics sector, leading to the establishment of the practice of prosthetics as a specialty and providing the first significant innovative impulse in prosthetic device design and production. In parallel, a number of technological and societal changes took place that did not immediately influence prosthetic design and practice, but which laid the groundwork for later significant changes in the sector.

The Great (US) Civil War Benefaction and the Emergence of the Prosthetic Specialist

The first of the wars, the American Civil War, left a large number of amputees in its wake and a recognition that these Veterans should be provided with some form of prosthetic

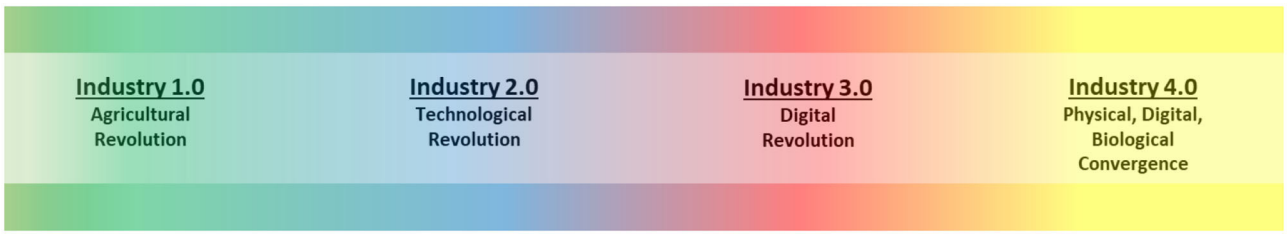

FIGURE 1 | Industrial innovation as a spectrum with no fixed points between eras.

device that they did not have to procure at their own expense. This led to a financial commitment by the US Government to provide all veterans with prosthetic devices through what was know as the Great Civil War Benefaction (11). This in turn generated a burst of prosthetic technology development activity, resulting in numerous patented prosthetic designs and represents the first time in history that "Industrial" thinking had been applied to creating solutions for what had previously been an *ad hoc* approach to prosthetic design and production. One of those inventors, an engineer, Civil War veteran and amputee, J.E. Hanger, not only patented innovative prosthetic designs, but also established the J.E. Hanger prosthetic workshops where employees specialized solely on making and providing prosthetic devices, as opposed to such devices being made by metalworking trades alongside other items such as horse shoes or tools and other implements. This was a pivotal shift in production and delivery model and with the ability to concentrate exclusively on the production of one type of device, standardized approaches and further design refinements began to emerge, leading to the establishment of prosthetics as a recognized, stand-alone trade.

Physical Therapy for Amputees and Modular Prosthetic Systems

The second of the wars, the First World War, created more than 41,000 amputees in Britain alone. In treating this large cohort, the first connections began to be made between amputation and the psychology of limb loss, changing how amputation and amputees were viewed in society (12). Physical therapy and gait training became part of the post-amputation recovery process for the first time. This wholistic approach to amputation led to an evolution of the construction and appearance of prosthetics limbs, including the reimagining of the prosthesis as a functional tool where terminal attachments resembling industrial equipment took the place of the prosthetic hand, as opposed to the prosthetic hand being an imperfect cosmetic replacement that provided some basic functions. The concept of the prosthesis as an industrial, functional tool disappeared by the 1950's, but foreshadowed the more radical reimagining of the prosthetic limb that was to come under Industry 4.0 (13).

In Germany the need for prosthetic devices for war amputees was also immense, sparking the development of a modular prosthetic system using standardized, mass produced components very much in the spirit of industrial revolution. Developed by the prosthetist Otto Bock, this innovation created a novel paradigm for prosthetic device production that assembled prostheses from components that had been designed, manufactured and quality tested and that were then attached to the socket, which was still hand-crafted. This allowed work processes to be rationalized, improving the efficiency of the prosthetist and providing a guarantee of safety and quality that is not possible to provide for hand crafted components (9). From this point onward, though still be done in practice, it was no longer *necessary* to hand craft prosthetic components other than the socket interface with the residual limb.

Early Human Movement Studies, Materials, and Electronic Innovations

While not applied to prosthetics at the time, developments in other sectors created innovative building blocks that would be applied to prosthetics technology and practices later, as Industry 2.0 gave way to Industry 3.0 in the 1960's.

One such building block was technology and processes which supported the study of human and animal movement. Interest in movement of the body dates back as far as the Renaissance, but it was the invention of the camera that allowed the first quantitative biomechanical studies to be done by Etienne-Jules Marey, using photos. Carlet and Muybridge followed, using early pressure recording shoes and film respectively (14). Over the course of Industry 2.0 these tools were refined and physiological monitoring technology such as VO_2Max was introduced. These were integrated into movement and gait analysis systems, which were then applied to the study of amputee gait and prosthetic device design beginning in the 1960's (15).

A second set of building blocks emerged in the chemical sciences from the 1870's to the 1930's, beginning with the study of natural resins and polymers and leading to the development of synthetically manufactured resins. From this a range of new materials, related processes and resulting products emerged (16). In the biomedical sciences, dentists were early adaptors of

[2]https://www.england.nhs.uk/commissioning/spec-services/npc-crg/group-d/d01/prosthetics-review/
[3]https://www.fda.gov/news-events/press-announcements/fda-approves-prosthetic-implant-above-knee-amputations
[4]https://www.3dsystems.com/learning-center/case-studies/bespoke-prosthetic-fairings-art-personalized-medicine

these materials during this period, using them for fillings and restorations. These materials did not begin to find their way into prosthetic limb production until the late stages of Industry 2.0 in the 1960's, when they were used to create both the outer cosmesis of prosthetic limbs and replacing hand carved wood sockets with composite laminated sockets (17).

A final example of building block innovation in this era is early work on myoelectric control, which began during the Second World War. The first known reference is a patent application for myoelectric control of a prosthetic hand made by Reinhold Reiter in Munich, Germany (18). This was followed by a rapidly growing body of research harnessing transistors to create upper limb myoelectric prostheses, work that was being pursued independently and collaboratively across the globe, including efforts in Japan, the US, Italy, Germany, Canada, UK, Russia, Sweden and Austria. These early digitally-controlled arms laid the groundwork for the use of micro-processors in upper limb, and later lower limb, prosthetic devices under Industry 3.0 (19).

INDUSTRY 3.0: THE DIGITAL REVOLUTION

Industry 3.0 is characterized by inventions such as the transistors, processors and computers which became smaller, more powerful and more flexible as the technology was refined and improved. Beginning sometime between 1950 and 1970 (3), and running until ∼2010, it ushered in a wide range of increasingly sophisticated prosthetic designs and components that were developed using interdisciplinary approaches. Human movement research was used, for the first time, to evaluate prosthetic design and function in a wide range of studies. This was a significant departure from previous prosthetic design practices where advances based on personal experiences were considered to be "trade secrets" to be passed down the generations and were not independently and objectively evaluated for function or effectiveness.

Digitization

Computers and digital technology became ubiquitous in workplaces and home during this era as desktop computers gave way to laptops, tablets and phones. This was supported by the development of cellular telecommunications, the Internet, Wi-Fi and Bluetooth, creating the ability for computing technology to follow the user wherever they went. Software was developed to run on the wide range of resulting hardware platforms, profoundly impacting the design and manufacturing sectors through tools such as Computer Aided Design and Manufacture (CAD-CAM), Flexible Manufacturing Systems (FMS) and Advanced Digital Manufacturing (ADM). This set the stage for disruption of prosthetic production under Industry 4.0.

Beginning in the 1980's digital design and manufacturing tools developed for other sectors began to make their way into the prosthetics sector. The application of CAD/CAM to the production of prosthetic devices was a natural offshoot of the success of CAD/CAM in other fields, adding efficiencies and reproducible accuracy. Early champions identified multiple benefits coming from the adoption of this technology in prosthetics (20). Seminal work was done to create sector specific software and hardware and included CANFIT (Vorum), Seattle ShapeMaker and CAPOD systems, with Vorum becoming an industry mainstay internationally (21).

Digitally supported advances were not limited to production processes. The far more visible and impactful digital transformation occurred in prosthetic componentry, setting new standards for prosthetic device function and end-user outcomes. Digital solutions first made their mark in upper limb prosthetics early in this period but it was in lower limb prosthetics that digital technology had the most perceptible impact on amputees, with the introduction of microprocessor controlled (MPC) joints. The Intelligent Knee (Blatchford Ltd., 1990) became the first MPC prosthetic knee to enter the marketplace quickly followed by the C-leg (Ottobock Gmbh, 1997) and then, MCP ankles and feet. These components profoundly changed the amputee experience by addressing functional needs and safety that previous prosthetic designs and technology were unable to (22–25).

Materials Sciences and Collaboration

Digital applications are the hallmark of Industry 3.0, but it was the adoption of synthetic polymers developed during Industry 2.0 most visually changed prosthetic technology in the early days of Industry 3.0. At that time prosthetic devices were still carved from wood, forged from metal and completed with customized leatherwork. New materials quickly transitioned the sector away from those materials to acrylic and polyester laminates. Though not "digital" in nature, the adoption of these new materials supported a paradigm shift that allowed novel prosthetic designs to be developed using structured processes incorporating interdisciplinary criteria, such as biomechanics and anatomy, into the design process (26). The move from carefully guarded "trade secrets" as the basis of prosthetic design to the use of objectively validated design iterations had begun.

Following World War 2, university research programs supporting improvements in prosthetic design were initiated. Universities began to influence the sector with developments such as the Supracondylar Socket, Patellar Tendon Bearing Socket, Four-Bar prosthetic knee joint mechanism, SACH (Solid Ankle Cushion Heel) Foot and Seattle Foot, each of which capitalized on materials sciences advances that were now coupled with structured engineering design practices and were carried out within an academic environment.

Collaborations between private industry and academic or public institutions also contributed, the classic example being the myoelectric arm combining the use of new materials with digital technology and developed by Ottobock Gmbh in collaboration with institutionally based research programs, such at the ones at I.N.A.I.L in Italy (18). A second example is the development of silicon liners as an alternative to cotton and wool stump socks and which provided additional benefits to amputees such as improved comfort and performance as well as providing suspension. Silicon liners, first developed in industry by Össur, were validated in scientific studies carried out at universities (27, 28).

Biomechanics Emerges as a Specialty and the Emergence of Clinical Outcome Measures

The objective study of prosthetic gait today cannot be imagined without digital technology. The integration of cameras capturing 3D movement with force sensors launched kinematic (motion) studies and kinetic (force) studies as a formal area of research. Leaders in applying this to pathological gait and prosthetic applications are Inman (UCLA-SF) and Perry (Rancho Los Amigos National Rehabilitation Center), the latter of whom expanded the tool kit by adding fine wire electromyography. Professor Paul aided, by Jarrett and Andrews, was instrumental in digitally integrating these tools, setting the stage for modern gait analysis systems (14, 29). These pioneering researchers built the foundation upon which laboratory-based research quantifying prosthetic gait is carried out to this day, allowing the examination of how prosthetic design influences amputee function and considering of how the resulting knowledge can be translated into clinical practice.

As the digital revolution gained momentum, the range of digital tools expanded to include scanners to capture residual limb shapes for modification in CAD/CAM systems, step count monitors allowing the tracking of community-based activity of prosthesis users (30), and tools to aid in objective alignment of prosthetic devices such as the 3D L.A.S.A.R. Posture (Ottobock), Compas and Smart Pyramid (Orthocare Innovations) (31, 32).

The development of these tools provided new perspectives on the prosthetic device provision process. Their objective, valid and reliable evidence-based outcome measures presented an alternative to the subjective "clinical expert" opinion that, up to that point, had been the standard for determining if prosthesis fit or function was satisfactory, or not. Despite their availability, uptake of the new digital tools was slow on the part of prosthetic practitioners in large part because they did not provide direct benefit to the clinical practitioner by improving efficiency, increasing productivity or boosting the bottom line. Compounding this, many of the new tools had high entry costs which proved to be a tangible barrier to adoption (33).

In the academic setting, the development of clinical outcome measures specific to prosthetics gained momentum in the mid 1990's. In particular, Gailey (University of Miami) and Hafner (University of Washington) were carried out critical, objective research on amputee gait that led to the development of a wide range objective clinical evaluation measures including, but not limited to, the AMPRO, AMPnoPRO (Gailey) and Plus-M (Hafner) (34, 35).

The need for such measures had already been identified in literature by Ramstrand and Brotkorb (36) but until the publication of the Levinson Report in 2011 (37), discussed in the next section, the audience for this growing body of research was limited to the academic setting. Prosthetic device providers still took much pride in their "hands on" experience-based knowledge often speaking of seeing "with their hands" during this latter phase of Industry 3.0 (9). Prosthetists have been slow to voluntarily adopt even simple clinical outcome measures for a range of reasons including a lack of the time it takes to carry the out an a lack of clarity as to the value they measures provide (38).

The entry costs for digital tools in this sector have significantly reduced over time but, in developing technology for this sector, this stage of prosthetic history illustrates the importance of balancing the full spectrum of economic costs vs. benefits; a critical factor in prosthetics due to the highly cost sensitive nature of the fee-for-device business model.

Adoption of Evidence-Based Practices and Clinical Outcome Measures in Clinical Prosthetic Practices

Clinical outcome measures and evidence-based practices were not seriously considered in the clinical setting until the publication of the aforementioned Levinson Report (37) by the US Department of Health & Human Services' Office of the Inspector General. The report was highly critical of prosthetic billing practices within the Medicare system in the United States, exposing a structural vulnerability in the sector, namely the lack of ability to demonstrate cost-benefit using objective criteria. It was a watershed moment, allowing insurers to require justification for reimbursement to be supported by objective, measurable outcomes as opposed to subjective expert clinical opinion or experience.

This report set in motion the translation of academic research on clinical outcome measures and digital evaluation tools practices into the clinical setting. Momentum was built by the active support for the development of evidence-based practices and tools through funding from organizations as diverse as: American Academy of Orthotists and Prosthetists (AAOP). American Orthotics and Prosthetics Association, the U.S. National Institutes of Health (NIH), U.S. Department of Defense (DoD) and the Orthotics & Prosthetics Education and Research Foundation (OPERF). This established an ongoing collaborative, interdisciplinary effort that is creating evidence-based knowledge on clinical and technical issues that relate to amputees and prosthetic care, including the use of digital tools.

INDUSTRY 4.0: PHYSICAL, DIGITAL, AND BIOLOGICAL CONVERGENCE

The term Industry 4.0 describes a convergence of Physical, Digital and Biological Systems that support the creation of "smart" technology or cyber-physical systems. The resulting technology can be networked and allows for the collection and storing of large amounts of data in data bases which have value in themselves as drivers of innovation.

Smart technology is often described as disruptive, spanning nine enabling technologies: Advanced Manufacturing, Additive Manufacturing, Data Analytics, Augmented Reality, Simulation, Horizontal/Vertical Integration, Cyber Security, Cloud Computing and the Industrial Internet.

At first glance this list has little relevance to the day to day production and provision of prosthetic devices but, as was the case in Industry 2.0 and 3.0, the prosthetics sector will

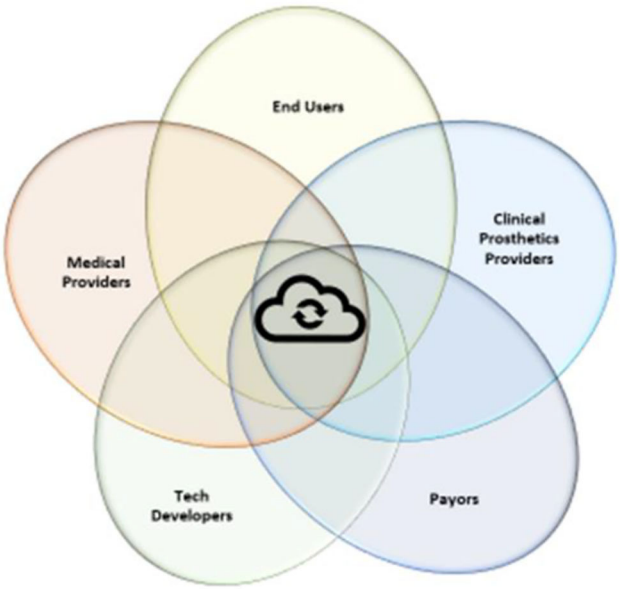

FIGURE 2 | Innovation will take place as various experts and bodies of knowledge co-create synergistically through the sharing of expertise and knowledge where they overlap with the other fields. The hub of this sharing will be data sharing.

be influenced and shaped by Industry 4.0. Technologies from this list that have already begun to manifest themselves in the prosthetics sector include Additive Manufacturing (3D printing), Smart Sensors, the Internet of Things (IoT), Blockchain, Software as a Service (SaaS), Machine Learning (M/L) and Big Data (39). In the prosthetics sector innovation will no longer be propelled by individual discoveries or events, such as a new material, but by overlapping influences that intersect and act synergistically at a data driven hub (**Figure 2**).

Socket Evolution: Materials Sciences and Machine Learning (M/L)

As in the previous industrial revolutions, advances in materials sciences are playing a prominent role in the evolution of prosthetic devices under Industry 4.0. In previous eras materials advances led to improved cosmesis, function and fit but the general form and design of prosthetic devices remained the same. Under Industry 4.0 data driven engineering approaches are being applied to materials development. This is allowing prosthetic designs metamorphosize and, in particular, there is now the ability to address shortcomings that have been identified with the classic prosthetic socket (40).

Changes in volume of the residual limb have been a long-identified problem with classic, rigid socket designs. Past approaches to managing residual limb volume fluctuation ranged from early adjustable sockets made of leather and lacers (41, 42) to adding and taking off stump socks over the course of the day, the latter of which is still done today but which is not a wholly satisfactory solution (43). Research has begun to point to the current standard of care being inadequate and thoughtful approaches are beginning to emerge (44). As in past eras, these draw on an expanding palette of materials alongside seeking more widely sourced innovative elements, for example the sports equipment and high-performance garment sectors and then combining them in more complex and novel ways. These adjustable socket solutions allow amputees to easily adjust socket volume manually over the course of a day eliminating the need to don and doff a prosthesis or to readjust laces, both of which take time and effort and require direct access to the prosthesis (45). Manually controlled designs have given amputees improved control over the fit of their sockets, but this is only the beginning. Early work on automated adjustable socket designs has started, creating a potential future where socket fit is adjusted automatically in close to real time (46). The added benefits of these engineering and technology-based developments are that they will allows for greater quality control in the production of the socket itself, reducing product liability risk and potentially creating efficiencies within the production and provision processes. This example is one which illustrates the increasing complexity of solutions and how ideas and components from across sectors can now be combined to become more powerful than in previous eras.

At the complex end of the spectrum tools such as 3D Printing will be supported by new processes for measuring residual limbs using smart phones, scanners and other imaging technology. Highly sophisticated and complex methods of objectively capturing surface anatomy, images of the underlying anatomical structures and potentially, pressure gradients (47) and tissue properties, will be combined with data bases of anthropometric and biomechanical measurements to which machine learning (M/L) will be applied and used to generate custom designed sockets (48, 49). 3D printing will allow integration of added value elements into final product, for example through the use of copper infused filaments with antimicrobial properties (50). The personalization of devices will be further informed by 3D motion data collected not only by researchers in prosthetics but also those from the physical and exercise therapy fields, such as that being used in the development of automated active assist devices to support rehabilitation (51). This is an interdisciplinary, wholistic re-imagining the prosthetic design and provision process and removes the last subjective step, the creation of a traditional socket, from the prosthetic production chain, making it theoretically possible for the entire prosthesis to be generated from objective design criteria using quality-controlled production methods. Fully automated socket design and production process may not ultimately be desirable as end-users will likely always wish to have and will benefit from having an expert assess prosthesis fit and function, but by using more data informed approaches in the creation of the socket the prosthetic provider will have a more objective baseline to begin an optimization process from.

Smart Garments and Smart Technology

Smart garments combine novel fibers and textiles with sensors and data streaming capabilities. They can be used to monitor and diagnose medical conditions or, by the very nature of

their properties, provide benefits to the wearer including equal or superior clinical outcomes, lower costs or better customer experience. Digital tools integrated into garments will become commonplace throughout the entire health care system and, in addition to enhancing care, will allow vast data bases to be created and mined to support decision making processes.

Smart materials have already entered the orthotics sector. Garments such as the Stoko Leggings and the DM Orthotics Scoliosis Garment harness properties created by the way a yarn is composed and spun or how the garment's components are combined in order to provide benefits that are novel or replace a more complex, and sometimes costlier, orthotic device (52). Smart garments will find their way into prosthetic designs as well, in the form of socket liner systems, control systems and clinical assessment tools. One garment, Hexoskin^{tm}, has been validated as a tool for collecting physiological measures for a range of activities (53), including walking and could conceivably be built into protocols for evaluating prosthetic function (54). Smart technology will allow sockets to become an active component of the overall prosthesis contributing to improved fit and function (55), much like the MPC knees and feet did when they first became available. "Smart" sockets will integrate sensors that monitor pressure, fit and temperature (56), and will eventually be able to respond independently and dynamically to an amputee's physiological state or activity, all whilst streaming collected data into data bases.

Socket liners will become active monitoring and data collection components that complement microprocessor-controlled components at the knee, ankle, elbow or wrist. Data bases resulting from smart sockets and liners will be cross referenced with existing anthropometric and biomechanics data bases to which machine learning (M/L) and generative design practices will be applied (57). This will support the development of components that allow more complex and natural movement (58) and which will integrate sensors that enable temperature, touch and pressure to be incorporated into local feedback loops. Other research, focusing on implantable neural interfaces and brain-controlled interface (BCI), aims to allow the integration of BCI into prosthetic designs to drive prosthetic component control systems and provide real-time neural feedback concurrently (59).

The prosthesis of the future will be one that is custom designed and produced for individual end-users using objective design tools and automated industrial production methods and will be fitted and maintained using smart tools that provide objective, close to real time, data. This will allow the prosthetist to complete the evolution from being a crafter and fitter of devices to become a clinical technology manager, in partnership with amputees.

Automation, Apps and Software-as-a-Services (SaaS)

In this transformation to becoming technology mangers, automation and Software-as-a-Service (SaaS) can assist prosthetists with improving quality and outcomes, whilst at the same time addressing productivity and labor shortages, another challenge currently faced by the sector and which will require the adoption of new approaches and technology.

The World Health Organization (WHO) identified a lack of skilled personnel at the international level. The National Committee on Orthotic and Prosthetic Education (NCOPE) has identified a labor shortage in the US (60), a prosthetic services review is underway in the National Health Service (NHS) in England (61), and the topic emerged anecdotally in Germany during data collection by Seibt in his study of how Industry 4.0 was changing the prosthetics sector (9). This is clearly a global sectoral challenge.

In sectors with labor shortages, including health care service delivery, automation processes and the use of AI can help ameliorate workload and productivity challenges and at the same time improve clinical outcomes (62). Automation as part of the solution has become a controversial one and is often met with fear and resistance, in particular where the transition to an "automated" data-based design and production process is being introduced to sectors that still engage in hands-on, craft-based production.

One of two perspectives on automation typically surface when discussing Industry 4.0. One provocatively presents the automated "robot" as a replacement for the worker. The other presents automation as a tool to help improve productivity and quality (63). The former narrative preys on the fear of change but is unlikely as robots and other forms of automation have limitations and are best used to replace repetitive and predictable tasks. They will become more flexible and applicable to a wider range of uses as their development matures and integrates artificial intelligence (AI), but it is highly unlikely, even in the long-term, that robots will replace the prosthetic clinician.

Efforts to move the prosthetics business model from fee-for-device to fee-for-clinical services have met with limited success globally. The prosthetics practitioner is increasingly challenged to find efficiencies within their current business model. This is where automation tools will be able to play a positive role, by allowing prosthetic practitioners to restructure their prosthetic design and production activities improve productivity. This is no a scenario in which robots take over. This shift will happen in parallel with software advances that improve the efficiency of clinic practices by re-shaping current administrative and business practices. Much of the focus in the prosthetics sector has been on hardware related innovations, but software innovation will have an equal impact. Two are worth exploring within the context of this paper, Apps and Software-as-a-Service (SaaS).

App is short for Application, a small piece of software designed to carry out a specific task using smart phones, tablets and other digital devices. Apps are already available in the prosthetics sector supporting tasks such as taking outcome measures, aligning, tuning or monitoring a prosthetic device, taking scans of body parts, interfacing with electronic medical records (EMRs) and providing a portal for communicating with payors (64). Apps are also empowering end-users by allowing amputees to self-manage their conditions across a range of situations from controlling MPC componentry to monitoring glucose levels. They can also support interactive and wholistic care models by enhancing communication and relationship

building with clients, an example being the Ottobock Gmbh. fitness app. Finally, apps can provide a portal to Software-as-a-Service (SaaS) to support communication and interactions with central fabrication facilities, assist with the product design and provision process and in facilitating business transactions.

SaaS consists of subscription-based software platforms that support a range of business, production and clinical activities. SaaS reduces the initial cost of purchasing software and the associated costs of maintaining software in house. It can assist a clinical practice to become leaner, aiding with billing and reporting to payors, communicating with clients, organizing clinical outcome measures and tracking quality assurance data. It can also support liability management processes and can help maintain business continuity in crisis situations. SaaS will become an essential component of the prosthetic business model of the future, working synergistically with the manufacturing software and hardware. Finally, SaaS's networked nature will allow for prosthetic clinics to network with others that use the same platform(s) in order to collaborate to create large data bases that can be shared and mined, helping network members to maintain a collective competitive edge.

If harnessed strategically, these tools can allow prosthetists to reduce the time they devote to either producing prosthetic devices or wrestling with administrative tasks, allowing them to reallocate that time to focusing on their relationships with their clients, the amputees.

Medical Treatment Advances

Industry 4.0 will also be accelerating advances in medical treatment options *via* the new tools being created and by using modeling, machine learning and AI to harness data bases. It is difficult to predicted what the impact will be as these developments are still in the early to mid stages of the innovation pipeline. But two examples with the potential to disrupt current prosthetic practices have matured and are translating to the clinical setting in developed nations.

The first is in the treatment of diabetes, which creates a high societal burden and is a common cause of amputation in developed nations (65, 66). A wide range of approaches to improving the management and treatment of diabetes is being pursued, crossing the spectrum from apps to allow diabetics to better manage their condition, for example by tracking what they are eating to medical device-based approaches such as continuous glucose monitoring and insulin pumps, which sit alongside surgical innovations such as islet transplantation. Finally, there is the emergence of personalized medicine approaches (67), with a large-scale Swedish study recently reporting improved health in Type 1 diabetics, including a 40% decline in amputation, when using a personalized medicine approach (68). These multimodal and technical advances are encouraging for diabetics and while reduced amputation rates indicate success, this will reduce the number of lower limb prostheses required in developed nations, which in turn will impact current prosthetic business models.

The second, specific to prosthetics, is osseointegration (OI) or bone anchored prostheses which sit directly on the intersection of the Physical, Digital and Biological as a medical-technological hybrid solution for existing amputees. OI involves attaching prosthetic leg componentry direct to a bone anchored implant, much like dental implants work. This eliminates the prosthetic socket completely and with it, many of the problems associated with the fit and use of sockets. OI provides the additional benefit of providing a secure interface between prosthesis and skeleton, which has been shown to improve osseo-perception and walking ability (69). It does not come without its own risks, such as implant loosening or failure or infection and skin irritation at the stoma, but international experience has shown it to be a viable option, even preferable, for some amputees having problems with socket fit (70). With the first FDA approval for use of OI in the US, significant resources are now being devoted to support key research centres internationally in reducing the risks associated with OI. It is expected that the use of OI will increase over the next decade, offering new possibilities for amputees as the benefits provided by OI are enhanced by more sophisticated, instrumented prosthetic technologies. OI is a classic example of Industry 4.0 embodying the physical, digital and biological in a single entity (31).

Customer Empowerment

Industry 4.0 is often presented using device-based, hardware and software examples. More difficult to quantify and express are the psychosocial changes occurring in this era, arising from the enthusiastic uptake of the concept of democratization of technology (71). The increase in access to information and communication technologies afforded by Industry 4.0 is supporting shifts in self-image and control, disrupting previous societal organization (72). Debate in society increasingly includes themes of self-empowerment with some persons with disabilities now striving to embrace themselves as they are or to articulate themselves clearly within society, as opposed to hiding their disability (73).

In health care, social media has shifted the power balance between patient and traditional health care provider (74). In the case of durable medical devices, including prostheses, social media has provided the users of prosthetic devices pathways to reach tech developers directly and *vice versa*. Amputees can now communicate their desires and selves directly to engineers and industrial designers, circumventing the traditional "clinical expert" filter who in the past formed a barrier between end-user and engineer (**Figure 3**).

The result is a range of new approaches to thinking about prosthetic function and design. Fairings are an example of this shift. An aesthetic innovation arising from industrial designers and engineers responding directly to consumer pull, fairings are a non-prescription, add-on product which allows significant personalization and styling of a prosthetic limb. Fairings give amputees the opportunity to express themselves in highly creative and personal ways at relatively low cost and are produced and suppled by new entrants in the prosthetic component sector who use tools such as 3D printing to create their products.

At the high-profile end of this newly created dynamic are social medial influencers such as Aimee Mullins, Amy Purdy and performance artist Viktoria Modesta. These publicly accessible voices speak about concepts such as the prosthesis as a

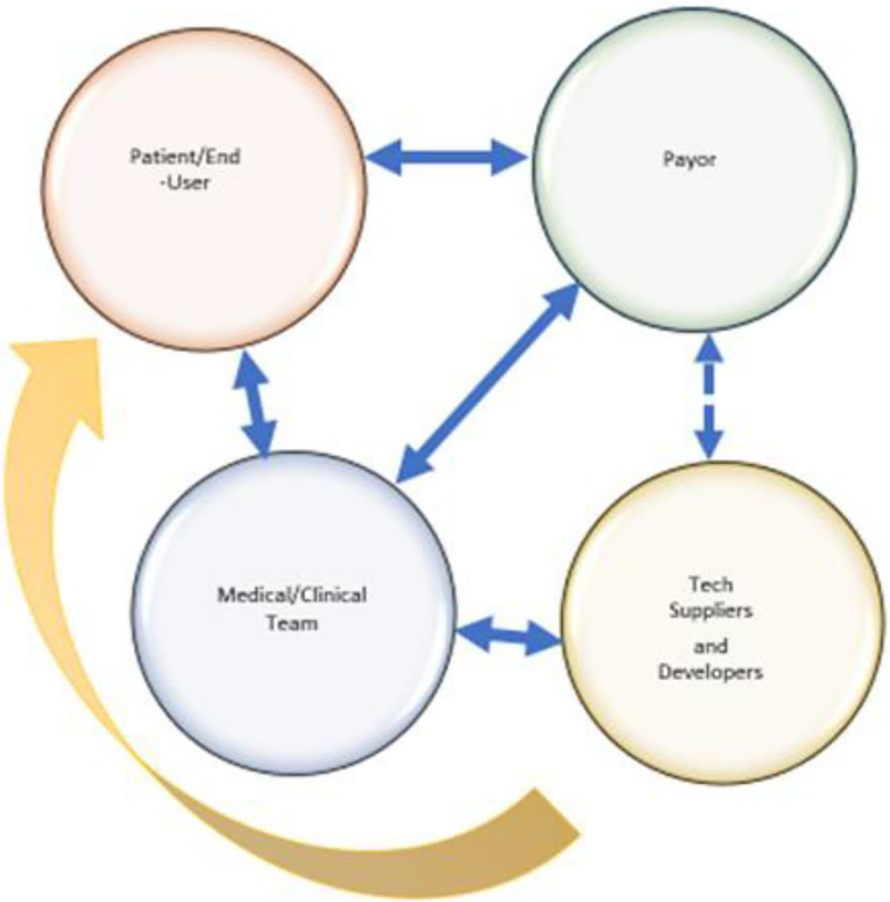

FIGURE 3 | Communication Lines: Established Traditional Communication Lines (Solid Blue), Intermittent Traditional Communication Lines (Dashed Blue), Social Media Enabled (Yellow).

functionally necessary accessory similar to eyeglasses, removing prostheses from the traditional "assistive device" category at the personal and conceptual level, and transforming it into a quasi consumer product. Under this paradigm a prosthesis is no longer is a device that attempts to "make whole," instead becoming a device which empowers and makes a statement. This approach is currently limited to those with the financial means to purchase multiple or artistically enhanced prostheses and will not necessarily be embraced globally across cultures, or by elderly amputees. It is nonetheless a change in how younger and more active people view both their disability, as well as their prosthesis.

At the other end of this dynamic are amputees who cannot afford any prosthetic device. This group found hope in the potential for additive manufacturing to allow them to gain access to simple, inexpensive prosthetic solutions, creating the first wave of open-source 3D printed hands (upper limb prostheses) that were designed, produced and supplied directly to the amputees, typically by engineering student volunteer groups. The ensuing demand and media attention made it clear that there was an unmet market need that was actively seeking a solution. This first iteration toward that solution did not upend the traditional prosthetic device market, but work in this space continues. A powerful characteristic of additive manufacturing is to allow fast, documentable design iteration and it can be anticipated that efforts by technology developers to create lower cost, customizable designs at the local level will eventually be successful and will have global impact in addressing this unmet market need.

The creation of lower cost, high quality, durable medical devices and health products supports more equitable health care options for all persons at the global level. In low resource settings, developments at the low-cost end of the innovation spectrum have the potential to not only reduce the cost of producing a device for those who cannot afford one, but also by making it possible to move the point of care to the local level. This is of critical importance as the expense of travel to a prosthetic clinic costs more than the device itself, creating an additional barrier. The cost of 3D printing technology continues to decree while the quality of prints increases, in parallel with smart phones become ubiquitous globally and provide access to telemedicine. The intersection of these trends will allow more sophisticated, mobile and affordable care to be delivered close to where it is needed.

This vision is consistent with other health care innovations for low resource settings which are now harnessing technology in this way (75).

Under Industry 4.0, low cost or a high level of convenience can no longer be equated with low quality or poor outcome. Disruption has already occurred in other health product categories who have adopted Industry 4.0 enabled approaches with some success. Hearing aids are one such product category, where technology is reconfiguring both the provision process and the business model. At the low-cost end of the market, the FDA has cleared the way for hearing assist devices to be available over the counter for those who cannot afford to, or find it inconvenient to, obtain a traditional hearing aid *via* a hearing aid clinic (76). The palette of options becomes even greater for those with hearing loss, as it is now also possible to do an on-line hearing assessment, to be supplied a hearing aid and to have that hearing aid fitted and tuned *via* web-based provision models. It is no longer necessary to physically go to a clinic for assessment and fitting. Orthodontic bracing systems are a second example. It is now possible to get a series of teeth aligners from storefronts or *via* web-based portals at a lower cost than traditional orthodontia (77). In person assessment followed by regular orthodontist appointments is no longer a necessity. This is not to say that traditional in-person clinical options have been replaced by online models. Many people will continue to prefer in-person models. However, these approaches and business models do provide new options of convenience for some and more equitable access for others. A decade ago both the technological and business approaches embodied in these two examples would have been unthinkable. Similar trends are likely to be seen in prosthetics as well. Indeed, the question must be asked: Why would the purchase of a prosthetic device be any less affordable, accessible, seamless or personalizable?

CONCLUSION

Dr. Stephen Seiler wrote: "History lectures are dangerous: one is forced to compromise completeness for the sake of flow and focus" when presenting a short history of endurance testing in athletes in 2011 (78). This paper does not presume to provide a complete history of technology development in prosthetics. Instead, it intends to identify congruences between the development of technology in society as a whole and advances in technology and practices in prosthetics. A short summary of progress in the prosthetics sector at the transition from one era to another is shown in **Figure 4**.

It has become commonplace to present technological advances within the framework of a series of Industrial Revolutions beginning in the mid 1700's as societies began to shift from being agriculturally based to industrially based. Over time, further Industrial Revolutions have been identified. Each is defined by a common theme and is discussed as a distinct era but has overlapping start and end points making it more accurate to think of this historical progression as a spectrum where the

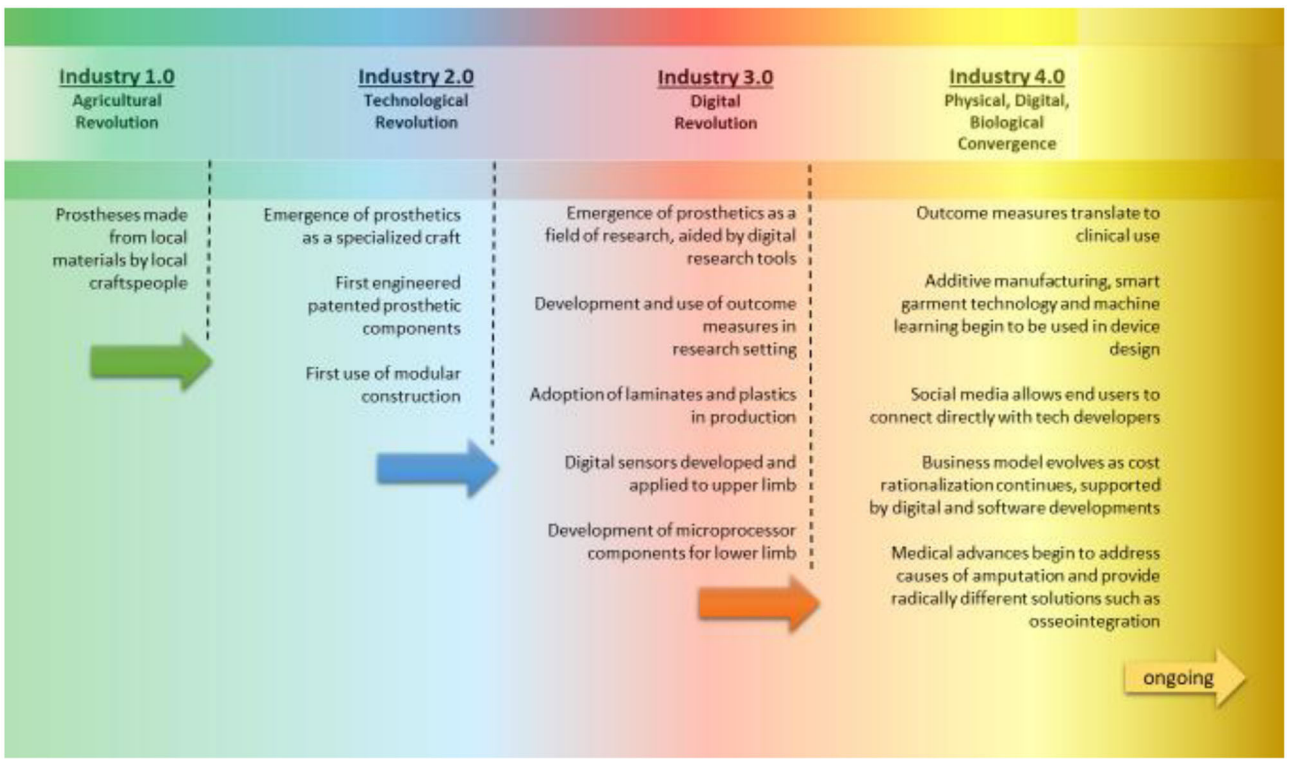

FIGURE 4 | Summary of prosthetic evolution highlights through the eras.

edges of each period blur into the next. In considering prosthetics using this framework we see that, like most other health sciences, it is a late adoptor of new technology and processes. Prosthetics is a sector that follows, not one that leads, which could explain some of the frustrations and business challenges faced by this relatively small field, when compared to other larger and nimbler sectors. It lacks the critical mass and resources required to take the risks associated with being a leader. At the same time this gives this sector a stability that is lacking in the volatile tech development world, which is of benefit to amputees for whom prosthetic devices are not the "latest gadget" but are critical to their ability to participate in life fully and productively.

The future cannot be predicted, but signposts indicate that prosthetics technology will continue to become more sophisticated, potentially crossing over with robotic or exoskeleton technology. Design and production processes will likely become more automated and will incorporate machine learning and artificial intelligence. With strategic shifts in thinking, Industry 4.0 could allow prosthetic providers to gain sufficient efficiencies within their fee-for-device business model to allow them to focus on providing their clinical services as technology managers, guiding and advising component choice, doing final fittings and ensuring that appropriate function is being provided. Two clear unknowns exist: One is the question of how the business of prosthetics will evolve to become more responsive to increasing consumer expectations while balancing that with payor limitations. The other is how advances in medical treatment options benefitting amputees, but potentially reducing the need for traditional prosthetic solutions, will change prosthetic services and role of the prosthetic provider. Relevance and viability in prosthetics, like all other health sectors, will require an openness to change and flexibility in approach in order for stakeholders to navigate this change in a sustainable way. If done smartly, it will benefits amputees globally. It will also allow prosthetic providers to re-imagine themselves and their role, ideally in a fulfilling way. So where does this leave the sector? In transition, as always.

AUTHOR'S NOTE

SR is an applied researcher with the British Columbia Institute of Technology MAKE+ group. She specializes in evaluation and product development projects in rehabilitation engineering, with a focus on prosthetics and orthotics. In 2013 she and collaborator, Dr. Michael Orendurff, Ph.D. won the Thranhardt Prize for their paper: "Can You Tell Which Foot is Which?." the first double-blind prosthetic foot evaluation that included community ambulation. She is currently involved in a diverse range of applied and academic research projects, including orthotic aspects of exoskeleton design, 3D printing innovation in foot orthotics, canine assistive devices and is a co-investigator on a study examining Glass Ceilings in Prosthetics and Orthotics. Her interests include how interdisciplinary innovation across a range of sectors is shaping prosthetic design and production practices and business models. She is Co-Editor-in-Chief of the Canadian Prosthetics and Orthotics Journal and Chair of the US Veterans Affairs Rehabilitation Research and Development (RR&D) Subcommittee on Rehabilitation Engineering and Prosthetics/Orthotics.

AUTHOR CONTRIBUTIONS

The author confirms being the sole contributor of this work and has approved it for publication.

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Characterizing the Gait of People With Different Types of Amputation and Prosthetic Components Through Multimodal Measurements: A Methodological Perspective

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Prosthetic gait implies the use of compensatory motor strategies, including alterations in gait biomechanics and adaptations in the neural control mechanisms adopted by the central nervous system. Despite the constant technological advancements in prostheses design that led to a reduction in compensatory movements and an increased acceptance by the users, a deep comprehension of the numerous factors that influence prosthetic gait is still needed. The quantitative prosthetic gait analysis is an essential step in the development of new and ergonomic devices and to optimize the rehabilitation therapies. Nevertheless, the assessment of prosthetic gait is still carried out by a heterogeneous variety of methodologies, and this limits the comparison of results from different studies, complicating the definition of shared and well-accepted guidelines among clinicians, therapists, physicians, and engineers. This perspective article starts from the results of a project funded by the Italian Worker's Compensation Authority (INAIL) that led to the generation of an extended dataset of measurements involving kinematic, kinetic, and electrophysiological recordings in subjects with different types of amputation and prosthetic components. By encompassing different studies published along the project activities, we discuss the specific information that can be extracted by different kinds of measurements, and we here provide a methodological perspective related to multimodal prosthetic gait assessment, highlighting how, for designing improved prostheses and more effective therapies for patients, it is of critical importance to analyze movement neural control and its mechanical actuation as a whole, without limiting the focus to one specific aspect.

Keywords: gait analysis, neuromechanics, prostheses, multimodal characterization, electromyograhy (EMG), muscle synergies, lower limb amputation

INTRODUCTION

The amputation of a lower limb is a complex and invasive surgery that is often needed due to traumatic events, vascular diseases, or tumors. The changes in demographics and the increasing incidence of the pathologies leading to an amputation will potentially impact the healthcare services, including the demand of prosthetic devices, and the number of persons living with the loss of a limb has been estimated to significantly increase in the next few decades (1). People with a lower limb amputation reported a quality of life (QoL) that is significantly lower with respect to the general population. Among others, factors such as the use of a specific prosthesis and close related factors such as residual stump pain and type of suspension were found to predict QoL scores significantly (2–6).

After the surgery, people with amputation must undergo a rehabilitation phase and a considerable walking training (7) to gain the ability to walk autonomously and safely with a prosthetic device (8, 9). People with mono-lateral amputation typically adopt a series of compensatory motor strategies involving the prosthetic side and the contralateral sound limb (10, 11), plus an increased involvement of pelvis and trunk (12–16). As a matter of fact, prosthetic gait reflects a mixture of deviations from normal gait and adaptive and compensatory motions dictated by residual limb functions. From a motor control standpoint, during the rehabilitation process, people with amputation must adapt their walking patterns to their new physical conditions, and this adaptation may result in changes in the way the central nervous system (CNS) controls the movement. Lower limb amputation leads to significant neural reorganization within the CNS, mostly due to the loss of the sensorimotor function caused by amputation (17) and to the new biomechanical condition induced by the type of amputation and by the used prosthetic device. The two factors influencing the gait in people with amputation are the level of the amputation (18, 19) and the type of prostheses (20–27). Regarding the former factor, the gait in people with transfemoral amputation (TFA) seems to be more asymmetric than that in people with transtibial amputation (TTA), with increased compensatory strategies, which, over time, may prove damaging to individuals (28). Concerning the latter factor, in recent years, the prostheses have improved in design, materials, and technology (29–32) to be more effective in terms of efficiency of ambulation, minimization of the asymmetries, and reduction of compensatory movements.

In this scenario, quantifying and characterizing the gait of persons with a prosthesis is an essential element to improve the development of new and ergonomic prosthetic devices, and to optimize the rehabilitation programs (33–36). Nevertheless, the heterogeneous variety of methodologies used to assess prosthetic gait limits the comparison of results from different studies and complicates the definition of shared guidelines. The quantitative prosthetic gait assessment should be focused on indicators of effective and ecological gait, such as the traditional gait parameters, level of gait asymmetries, metabolic consumption, and the amount of compensatory muscle activation, but the assessment of such an heterogeneous scenario requires novel research methodologies (37–41).

From this standpoint, the adoption of a multimodal approach is needed for a proper prosthetic gait evaluation. The project "Modular motor control of the contralateral sound limb in people with lower limb amputation: neuromechanical assessment of the prosthetic components in the control of locomotion," funded by the Italian Worker's Compensation Authority (INAIL), led to the generation of an extended dataset comprising multimodal measurements involving all the common gait analysis instruments. By discussing the results of different studies published within the project, identifying some peculiarities in the used instrumentation and highlighting the importance of the related indices, we here provide a methodological perspective related to multimodal prosthetic gait assessment.

Starting from the dataset recorded during the project activities in the first section, we then describe the results obtained in five different studies, published along the project activities that analyze specific aspects of gait of people with amputation. Each study yields both direct evidence, coming from the recorded data and based on the specific indices used to quantify the gait performance, and indirect evidence emerging from the interpretation of the results. These direct evidence and indirect interpretations are then summed up and integrated in the final perspective section of the article, where we provide a methodological perspective supporting the importance of multimodal prosthetic gait assessment.

POPULATION, MEASUREMENTS, AND PROTOCOL

The population enrolled during the project activities underwent a typical gait analysis protocol executed with a multimodal set of measurements in terms of instrumentation and variable number of subjects, as reported in **Table 1**. This perspective article takes into account the results of 5 different studies. All the experiments were carried out at the Rome Branch of Prosthetics Center of INAIL, at the CTO Andrea Alesini hospital of Rome.

In total, 57 recordings from subjects with unilateral TFA and 20 recordings from subjects with unilateral TTA were performed. The subjects with TFA wore three different types of prostheses: mechanical prosthesis (TFA_M) and two types of prostheses with microprocessor-controlled knees (MPKs), namely C-Leg (TFA_C) and Genium (TFA_G) prosthesis (Ottobock, Duderstadt, Germany). All subjects with TFA and TTA were provided with the same type of prosthetic foot (Ossur Variflex, Reykjavik, Iceland), whereas the socket was custom-made and adapted to the single user needs before the gait analysis protocol by an experienced physician. All subjects with lower limb amputation were experienced prosthesis users (i.e., able to walk safely with a prosthetic device for more than 2 years). In addition to the TFA and TTA populations, 40 healthy subjects were recruited as the control group (C), and they were age–sex–speed matched with the amputees group.

Walking tests at a self-selected comfortable speed were performed on a 9-m long walkway instrumented with two force platforms (Kistler9286AA, Winterthur, Switzerland). Control subjects were requested to walk also at a lower speed to match

TABLE 1 | Populations characteristics across the 5 studies.

	Without EMG			With EMG	
Population age height mass	**Varrecchia et al. (42)**	**Castiglia et al. (43)**	**Ranaldi et al. (44)**	**De Marchis et al. (45)**	**Tatarelli et al. (46)**
TFA_M	*n* = 9 (1 F) 56.9 ± 12.6 yo 169.5 ± 4.9 cm 79.8 ± 16.8 kg	*n* = 9 (1 F) 56.9 ± 12.6 yo 169.5 ± 4.9 cm 79.8 ± 16.8 kg			*n* = 10 (1 F) 58.7 ± 11.5 yo 171.3 ± 7.0 cm 79.7 ± 15.9 kg
TFA_C	*n* = 17 (2 F) 58.2 ± 14.0 yo 172.3 ± 7.4 cm 83.0 ± 13.1 kg	*n* = 17 (2 F) 58.2 ± 14.0 yo 172.3 ± 7.4 cm 83.0 ± 13.1 kg	*n* = 7 54.6 ± 14.6 yo 173.4 ± 6.4 cm 84.4 ± 16.9 kg	*n* = 7 54.6 ± 14.6 yo 173.4 ± 6.4 cm 84.4 ± 16.9 kg	*n* = 16 (2 F) 56.4 ± 14.1 yo 172.4 ± 6.9 cm 81.7 ± 13.5 kg
TFA_G	*n* = 14 50.2 ± 12.8 yo 177.5 ± 16.0 cm 88.0 ± 13.0 kg	*n* = 14 50.2 ± 12.8 yo 177.5 ± 16.0 cm 88.0 ± 13.0 kg	*n* = 7 46.8 ± 14.5 yo 177.0 ± 7.2 cm 87.0 ± 13.8 kg	*n* = 7 46.8 ± 14.5 yo 177.0 ± 7.2 cm 87.0 ± 13.8 kg	*n* = 11 48.4 ± 13.5 yo 177.5 ± 5.8 cm 84.6 ± 12.2 kg
TTA	*n* = 15 52.8 ± 14.5 yo 176.4 ± 5.4 cm 87.4 ± 11.1 kg				*n* = 11 59.4 ± 12.8 yo 176.0 ± 6.2 cm 85.1 ± 12.7 kg
C	*n* = 40 (3 F) 54.9 ± 12.3 yo 172.8 ± 7.9 cm 83.5 ± 15.7 kg		*n* = 12 53.6 ± 8.1 yo 176.9 ± 7.0 cm 78.2 ± 6.6 kg	*n* = 12 53.6 ± 8.1 yo 176.9 ± 7.0 cm 78.2 ± 6.6 kg	*n* = 22 (3 F) 52.8 ± 14.5 yo 176.4 ± 5.4 cm 87.4 ± 11.1 kg

the TFA and TTA groups. A six-infrared camera optoelectronic motion analysis system (SMART-DX 6000 System, BTS, Milan, Italy) was used, with passive spherical markers placed according to a modified Davis' protocol (47). In subjects with TTA and TFA, the amputated limb markers were placed over symmetrical points with respect to the homologous marker's position on the non-amputated limb. Electromyographic (EMG) signals were recorded using a wireless system (FreeEMG 1000 System, BTS, Milan, Italy). Muscle activity was recorded from 12 muscles of the sound side (right side for the controls).

RESULTS

The results coming from the multimodal analysis underlying this paper are briefly reported in **Table 2** in terms of their direct and indirect interpretations. The following paragraphs report details on the mentioned studies that will serve as a base for the final perspective about the emergent features of prosthetic gait that are shown by adopting this kind of analytical approach.

Kinematic, Kinetic, and Energy Consumption Patterns

By including two different amputation levels (i.e., TFA and TTA) and three different types of prostheses for the TFA (i.e., mechanical, C-Leg, and Genium), a comparison of spatiotemporal parameters, plus kinematic and kinetic indicators, as compared to a speed-matched control group, was conducted in Varrecchia et al. (42).

The study highlighted that some patterns characterize prosthetic gait in general, regardless of the type of amputation and the kind of used prosthesis, whereas the others are specific for TFA and are dependent on the type of prosthetic knee ($I\text{-}A_{DIR}$).

From a purely kinematic standpoint, TFA and TTA subjects show an increased step width, step length, and double support duration. These subjects also show an increased pelvic obliquity and a higher range of motion (RoM) in trunk movements when compared with controls, regardless of whether the leading limb was the prosthetic one or the sound one, indicating that most of the compensation happens through the pelvis and trunk ($I\text{-}B_{DIR}$). An increased stance/swing ratio characterizes the sound side. From a kinetic standpoint, the prosthetic gait is characterized by an increased initial peak in the ground reaction force (GRF) on the sound side. All these alterations might be due to a lack of sensory feedback and to an absence of perception regarding foot placement ($I\text{-}B_{IND}$).

However, besides these common alterations in gait patterns, some additional changes characterize the gait of people with TFA. Kinematic alterations include a reduced stance/swing ratio in the prosthetic side and a higher hip and knee RoM in the sound side. Kinetic alterations include an increased initial peak in the GRF on the prosthetic side, suggesting that TFA are not able to generate adequate forces during stance ($I\text{-}B_{IND}$).

When considering the effect of the device, subjects using a Genium prosthesis have a lower pelvic obliquity when compared to TFA_M, a higher hip and knee RoM on the prosthetic side and an increased step length when the sound limb leads. This might indicate that more advanced knee prostheses have a general better performance in gait ($I\text{-}A_{IND}$), leading to a reduced compensatory effort ($I\text{-}C_{IND}$).

Since the main alterations are present in the TFA gait, the potentially induced increase in the metabolic consumption could

TABLE 2 | Direct and indirect evidence and interpretations emerging from the findings of 5 different studies.

	A-Spatiotemporal		B-Kinematic/Kinetic		C-Energy consumption		D-Motor control	
	Direct	**Indirect**	**Direct**	**Indirect**	**Direct**	**Indirect**	**Direct**	**Indirect**
I-Varrecchia et al. (42)	Some patterns are typical of TFA gait	More advanced prosthetic knees perform better in gait	Compensatory mechanisms happen through pelvis and trunk	Compensation of lack of sensory feedback and foot placement TFA unable to generate adequate forces during stance		Less advanced prosthetic knees need higher compensatory effort		
II-Castiglia et al. (43)			Normalization of pelvic obliquity on the prosthetic side of the subjects using more advanced prosthetic knees	Pelvic obliquity is related to "hip hiking" strategy of the affected side Pelvic obliquity allows limb forward progression during walking	Pelvic obliquity affects energy recovery	More advanced prosthetic knees can reduce the risk of low back pain		
III-Tatarelli et al. (46)				Increased coactivation reflects the compensatory increase in stiffness and changes in force production capacity Compensatory coactivation of the sound limb muscles may relevantly contribute to asymmetry		Compensatory coactivation of the sound limb muscles may relevantly contribute to excessive energy expenditure		The most critical phases in prosthetic gait are the double support ones Differences in coactivation depend more on the inertial properties of the prostheses rather than control mechanisms
IV- De Marchis et al. (45)				Same synergies between TFA and controls indicate same biomechanical functions		Synergy activation modifications during weight transfers represent an efficient compensatory mechanism	Motor coordination schemes in TFA are not different from the case of non-pathological gait	The most critical phase in TFA gait is the weight transfer phase from the sound limb to the prosthetic one. Alterations in synergy recruitment constitute a speed independent marker of TFA gait
V-Ranaldi et al. (44)		Principal components of elevation angles might be related with the spatiotemporal gait parameters						Double support phases are the most critical to be managed in prosthetic gait Alterations in principal components characteristics are related to altered neuromuscular control strategy

be explained by an energy-related indicator able to discriminate among different types of prostheses, shedding light onto the efficacy of different prosthetic components. Compared with speed-matched healthy controls, subjects with TFA, indeed, are characterized by a lower ability to recover mechanical energy at each walking step, regardless of the type of prostheses. Among the various spatiotemporal and kinematic modifications in the TFA gait, the only variable that is related to the lower energy recovery is the pelvic obliquity on the prosthetic side (43) (II-C_{DIR}). The pelvic obliquity is the rotation of the pelvis around the coronal plane, defined as the angle between the horizontal plane and the mediolateral axis of the pelvis, and its increase has been shown to be a compensation strategy used to propel the limb and recover energy (II-B_{IND}). This parameter not only correlates with relevant energy consumption measurements, but also highlighted that most advanced technological prostheses, such as Genium, likely require less compensation in the pelvic obliquity to recover the same amount of energy at each walking step (II-B_{DIR}), thus potentially reducing the risk of low back pain (II-C_{IND}).

Electrophysiological Features of Prosthetic Gait

Although these considerations allow us to better understand the effect of prosthetic gait on the movement mechanical outcome, yet the causes leading to such modifications can only be estimated, and the underlying changes in the control strategies adopted by the neuromuscular system can be roughly inferred but cannot be quantitatively accessed. A multimuscle activity measurement involving the sound limb has been performed to further advance our comprehension on the underlying neuromuscular strategies, by recording the EMG activity of 12 mono- and bi-articular muscles acting at the ankle, knee, and hip joints.

To better understand the coordination mechanisms of such muscles, a compact indicator, consisting of a time-varying function, has been used to describe the global neuromuscular strategy adopted by a subject in modulating the simultaneous activation/deactivation of many muscles during gait (48). The analysis on a population of TTA, TFA, and controls highlighted that people with amputation had a coactivation profile similar to the control population. However, the prosthetic gait led to an increased level of simultaneous activation during the loading response and push-off phases, whereas this coactivation was decreased during midstance and swing (46). This increased coactivation probably plays a role in the prosthetic gait asymmetry and altered energy consumption (III-C_{IND}). Among people with TFA, the used kind of prosthesis had an effect on the global coactivation, as it resulted lower in C-Leg users when compared with Genium and mechanical prostheses users. In detail, the increased coactivation levels that are recorded during the general prosthetic gait can be seen as a cause for the decreased force generation capacity and as an additional source of asymmetry (III-B_{IND}). Moreover, the same coactivation can also be seen as an important feature of motor control, isolating the double support phases of gait as the most critical for walking with a prosthesis, in which the different inertia properties of a prosthetic leg with respect to the intact limb might play a key role (III-D_{IND}).

Neuromechanics and Motor Control

The aforementioned multi-muscle EMG recording can also take advantage of the nowadays widespread and clinically relevant theory of modularity in motor control (49). By using the synchronous muscle synergy model, it was possible to identify low-dimensional control structures characterizing the muscle coordination of TFA subjects. In De Marchis et al. (45), it was shown that, despite the visible alterations in muscle activity, the complexity in muscle coordination did not change, as the TFA group exhibited 4 modules, which is the same number of muscle synergies typically expressed by control populations (IV-D_{DIR}). When analyzing the spatial structure of these modules (i.e., the groups of muscles working synergistically), it was shown that it is shared between TFA and controls, consisting of a weight acceptance module at sound limb heel strike, a propulsion module before toe-off, a swing module, and a late swing deceleration module before heel strike. This indicates that the main underlying biomechanical functions were preserved (IV-B_{IND}). However, the difference between TFA and controls was clearly visible in the activation of two out of the four identified modules: a significantly prolonged activation of the propulsion module (calf muscles) and an abnormal activation of the late swing deceleration module (hamstring muscles) during the second double support phase with respect to speed-matched controls (IV-D_{IND}). This result indicates that the most critical phase in gait of people with TFA is the second double support phase, corresponding to the weight transfer from the sound limb to the prosthetic one (IV-D_{IND}), potentially reflecting an efficient compensatory mechanism that enforces the interpretation of the results on the coactivation strategies (IV-C_{IND}).

Analogous results can be obtained by exploiting the planar covariation law of elevation angles. Following the same rationale adopted for the muscle synergy analysis, it is possible to define a common spatial organization for the behavior of the elevation angles of the three lower limb segments (50). With this description, both limbs of the patients and healthy subjects share the same covariation domain, with differences that are limited to the trajectories of the three angles in this space (44) (V-D_{IND}). Coherently with all the other studies presented before, most of the differences are to be ascribed to the management of the body weight and on the contact phase of the limb with the ground (i.e., the stance phase), with the prosthetic limb showing a higher degree of correlation among the three leg segments, as a direct consequence of the control mechanisms of the prosthetic knee (V-D_{IND}). Moreover, the planar covariation law of elevation angles is a compact description that directly approximates the kinematics of the two legs; as a consequence, it is ideally possible to exploit this economic description of gait for predicting different quantitative measures of walking behavior, such as the spatiotemporal parameters, giving rise to a variety of applications for prosthetic control and rehabilitation (V-A_{IND}).

DISCUSSION

Previous studies have highlighted some important features of prosthetic gait, including stability, spatiotemporal gait parameters, and symmetry, gathering relevant information by using a reduced set of sensors (51–57). However, due to the heterogeneous nature of the prosthetic components, type of rehabilitation, experience with prosthesis use, and the amputation characteristics themselves, a multimodal approach to such gait analysis could be able to shed light on some important features related to these multiple factors (38, 58, 59), thus supporting the clinical practice (60–62). The neuromechanical analysis of gait, bringing together the information on biomechanical aspects and neural control aspects, takes advantage of this multimodal approach. This adds to the lack of studies including an EMG analysis of the contralateral limb, which provides a powerful insight into how the CNS is adapting to walking with a prosthesis (63–65), in addition to the changes appearing at the level of the residual musculature (66, 67). Overall, kinematic, kinetic, and surface EMG gait findings reflect the compensatory efforts developed by people with amputation to protect the soft tissues of the prosthetic limb and to deal with the new prosthetic limb condition.

Within the framework of the INAIL-funded project activities, the aim of this perspective is to fill the gap between the complexity of prosthetic gait and the necessity of a complete set of measurements. The summary of the discussions of the outcomes of the project presented here, reported in **Table 2**, highlights how different analyses can yield a wide overview of the characteristics of the prosthetic gait, reducing the number of indirect considerations (i.e., speculations) that are needed to describe all the factors starting from the results of an incomplete set of analyses. **Table 2** reports, for each of the five studies, both the direct evidence, as obtained from the analysis of specific quantities of prosthetic gait, and the indirect interpretations related to aspects that are not explicitly considered in the analysis. These are discussed in the following sections.

Spatiotemporal, Kinematic, Kinetic Metabolic, and Motor Control Aspects of Prosthetic Gait

From a spatiotemporal parameters point of view, a typical gait analysis can be combined with the analysis of the coordination of the lower limb segments, linking concepts related to the biomechanics to motor control investigations. Although the analysis of the spatiotemporal parameters of gait is well-established in the scientific literature, it leads to results that are not directly linked to how the movement is controlled, and thus might fail in identifying the key adaptation mechanisms that underlie walking with a prosthesis and the variables relevant to patient's satisfaction (68). In this scenario, the methodological approaches from I (42) and V (44) could take a strong advantage from a joint analysis, with important implications in engineering (e.g., development of advanced control systems for prostheses based on the underlying motor control mechanisms) and clinical practice (e.g., the use of motor control theories as a benchmark for functional gait recovery).

When dealing with kinematic and kinetic analysis of gait, the multimodal approach is of critical importance for a correct interpretation of the results. This happens with the interpretation of the results of both standard gait and coactivation analyses, from which it is possible to only indirectly suppose that prosthetic gait is characterized by a lower capacity of force generation, whereas the combination of the two studies reinforces this hypothesis. These considerations suggest that the methodologies of I (42) and III (46) have an important complementary role in the assessment of prosthetic gait dynamics.

Moreover, the synergy analysis confirms that the same biomechanical functions of physiological gait are preserved but controlled with different timings; by combining these results with those on pelvic obliquity and general kinematics, it is possible to understand how the abnormal activations in the synergy profiles reflect on the altered movement biomechanics. In the same manner, both the discussions about motor control and energy consumption can be summarized by focusing on the difficulties and asymmetries in the management of the weight shift phases (at the beginning and the end of the stance phase of both legs); this, combined with the results on the pelvic obliquity and with the characterization of the energetic inefficiency, confirms in a quantitative way the already published results, that identify the double support phases of prosthetic gait as the most critical, both from a stability and an energetic point of view. These considerations highlight the importance of connecting the multi-muscle EMG measurement and synergy analysis of IV (45) with the methodologies used in II (43) related to the body center of mass and to pelvic kinematics, for a complete understanding of the interplay between compensation mechanisms and energy consumption.

Perspective of the Multimodal Analysis of Prosthetic Gait

Future studies should explore whether the adoption of a multimodal approach can capture the alterations in performance-based walking measures (69), the metabolic cost of walking (32, 70), self-perceived mobility and balance outcomes (37, 71), and the acceptance of prostheses (3). In the clinical practice, the outcomes of the rehabilitation therapies are often measured by means of qualitative scales, such as the K-Level, which to date is the main scale used by physicians to choose the most adequate prosthetic device. Some semi-quantitative scales have also been proposed, like the Amputee Mobility Predictor (AMP) scale (72), in which some spatiotemporal parameters of gait are used to refine the information provided by the K-level classification. In addition to these indices, several clinical scales describing the patient's QoL are adopted as a description of the follow-up of the therapies, such as the Amputee Activity Survey or the 12-min walking test (73). In general, all the current clinical scales are pseudo-subjective, based on questionnaires that are dependent on the personal perception of either the physician or the patient. Nevertheless, they might fail in providing insight into the interplay between the prosthesis and the patient so that

the adopted solutions might not be always optimal. Although it remains still feasible to use those measures in the clinical practice, the recent advancement in prosthesis technology could make the patients reach high scores for most of these scales. In this sense, the definition of more objective scales, exploiting engineering tools, can help in reaching a higher level of understanding of how the choices of the prosthetic component affect the movement control; this can lead to the development of prosthetic devices that reach a higher degree of integration with the subject's motor control strategies.

CONCLUSIONS

The meta-discussion presented here was elicited by the heterogeneous framework of results obtained within the project "Modular motor control of the contralateral sound limb in people with lower limb amputation: neuromechanical assessment of the prosthetic components in the control of locomotion" funded by INAIL. All the considerations from the related studies strongly highlight the importance of applying a multimodal approach when analyzing gait in people with a lower limb amputation; as a matter of fact, despite the huge scientific effort of the last two decades, this condition is still partially unknown to date, and the compensations that are necessary for reaching stable gait with a prosthesis are highly complex and cannot be characterized as a whole without a complete recording and analysis of all the influence factors. Consequently, this strongly recommends that, for designing improved prosthetic device, develop more advanced and physiologically inspired prosthesis control systems, and plan more effective therapies for these patients, it is of critical importance to analyze movement neural control and mechanical actuation as a whole, without limiting the focus to one specific aspect.

DATA AVAILABILITY STATEMENT

The original contributions presented in the study are included in the article/supplementary material, further inquiries can be directed to the corresponding author.

ETHICS STATEMENT

Ethical approval was not provided for this study on human participants because it analyses previously published results. Details on ethics can be found in the original referenced articles. The patients/participants provided their written informed consent to participate in this study.

AUTHOR CONTRIBUTIONS

CD, SR, and SFC conceptualized the study. CD and SR prepared the original draft of the manuscript. SC, AR, MS, FD, and FL provided resources and administered the project. All authors reviewed and edited the manuscript for important intellectual content.

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Functional Mobility Training With a Powered Knee and Ankle Prosthesis

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Limb loss at the transfemoral or knee disarticulation level results in a significant decrease in mobility. Powered lower limb prostheses have the potential to provide increased functional mobility and return individuals to activities of daily living that are limited due to their amputation. Providing power at the knee and/or ankle, new and innovative training is required for the amputee and the clinician to understand the capabilities of these advanced devices. This protocol for functional mobility training with a powered knee and ankle prosthesis was developed while training 30 participants with a unilateral transfemoral or knee disarticulation amputation at a nationally ranked physical medicine and rehabilitation research hospital. Participants received instruction for level-ground walking, stair climbing, incline walking, and sit-to-stand transitions. A therapist provided specific training for each mode including verbal, visual, and tactile cueing along with patient education on the functionality of the device. The primary outcome measure was the ability of each participant to demonstrate independence with walking and sit-to-stand transitions along with modified independence for stair climbing and incline walking due to the use of a handrail. Every individual was successful in comfortable ambulation of level-ground walking and 27 out of 30 were successful in all other functional modes after participating in 1–3 sessions of 1–2 h in length (3 terminated their participation before attempting all activities). As these prosthetic devices continue to advance, therapy techniques must advance as well, and this paper serves as education on new training techniques that can provide amputees with the best possible tools to take advantage of these powered devices to achieve their desired clinical outcomes.

Keywords: physical therapy, above-knee amputation, ambulation, robotic prosthesis, rehabilitation, artificial leg, prosthesis training, transfemoral amputation

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INTRODUCTION

Individuals with lower-limb loss at the transfemoral or knee disarticulation level lose a significant amount of mobility due to missing both an ankle and knee joint (1–3). Various daily activities are affected including walking in the community, negotiating obstacles within their home or work environment, navigating curbs or ramps, and transitioning to and from a seated position. These routine tasks are very challenging for most individuals because commercially available prosthetic joints are mechanically passive devices and cannot provide joint power similar to anatomical joints. During walking, passive ankle joints cannot provide ankle push-off power in a late stance and passive knees cannot actively extend during the swing phase, thus, requiring individuals to provide

active hip flexion to advance the prosthetic limb. While ascending steps, ascending an incline, or standing up, unilateral transfemoral amputees must rely heavily on their non-amputated limb and may resort to other compensatory movements such as increased upper extremity support (4, 5) or excessive vaulting (6, 7). As a result, these individuals expend more energy and have slower walking speeds than persons without an amputation (8, 9).

Prosthetic knees and ankles that have the capability of restoring power at the missing joints are becoming commercially available (10, 11) and with several more in development (12–20). The availability of power can allow for more normative gait kinematics (15, 21, 22) and can re-introduce an individual to activities they may not have completed since prior to their amputation including climbing stairs in a reciprocal manner (23–26), ambulating up and down, long and/or steep inclines with confidence (25, 27, 28), and transitioning to standing with more equal weight bearing (29, 30). The addition of power at the knee joint may reduce the occurrence of overuse injuries that occur after long-term prosthesis use. Providing the user with active powered plantarflexion during ambulation has the potential to decrease hip joint effort on the prosthetic side (31). An overarching goal for powered devices is to utilize both lower extremities more equally for daily activities. Achieving this goal may allow users to prevent further musculoskeletal injuries, as well as improve their strength, balance, and postural stability.

While the promise of powered prostheses is abundant, clinical instruction and training are needed to provide amputees with the best outcomes possible and enable them to utilize these devices to their fullest potential. Rehabilitation of transfemoral amputees varies greatly among therapists and rehabilitation facilities (32). This variation may be due to differences in patient populations across facilities, etiology, availability of therapy equipment, clinical skills and education of staff on the various devices, and accessibility to prosthetists and manufacturers for extended device training. While active devices can restore additional functional activities (e.g., reciprocal stair climbing), learning to incorporate all these features and optimize ambulation requires user and clinician education.

Educational materials are necessary to instruct patients on how to properly use these devices. For example, the manufacturer suggested training techniques for the Ossur Power Knee are divided into levels: initial training includes walking mode, intermediate training includes stair and ramp descent modes, and advanced training includes stair ascent mode (10). One Power Knee study cited 16 h of training to allow users to accomplish sit to stand transitions, stair ascent/descent, incline walking, and walking over uneven terrain (24), while another study showed that transfemoral amputees first fit to the Ossur Power Knee achieved functional mobility milestones in less time than those who were fitted first to a non-powered knee devices (16, 33). Studies investigating transtibial amputees' functional mobility during incline walking and stair climbing with the emPOWER (formerly the BiOM)-powered ankle indicated that more focused and device-specific gait training is recommended (31, 34, 35). Introducing clinicians to these devices with more opportunities for appropriate education and training will likely have a positive impact on physical therapy practice, goal setting, compliance of wear, and use of advanced devices for patients with transfemoral amputations (36).

The purpose of this paper is to help fill the gap in education regarding instructing transfemoral amputees on the use and functionality of a powered knee and ankle prosthesis. These techniques and tools were developed during the training of thirty transfemoral and knee disarticulation amputee users over the last 10 years at a nationally ranked physical medicine and rehabilitation research hospital. The training was designed to meet the goals of independent ambulation through all functional mobility modes including level-ground walking, incline walking, stair ascent and descent, and sit-to-stand and stand-to-sit transitions within a rehabilitation setting.

METHODS

Thirty patients (**Table 1**) who have had a unilateral transfemoral or knee disarticulation amputation participated. All patients provided written informed consent as approved by the Northwestern University Institutional Review Board. Individuals were independent in ambulation for level-ground, inclines, and stairs with their current device and classified as varied cadence community ambulators (centers for Medicare & Medicaid Services K3 and K4 level). Each subject was evaluated by a certified prosthetist and either used their clinically prescribed socket or duplication of their home socket for use during training sessions. If necessary, and often due to the added weight of the device, adjustments were made to the suspension of the device either by socket modifications or the addition of socks or a suspension belt.

Powered Knee and Ankle Prosthesis Description

The powered knee and ankle prosthesis (15) initially used for this study were designed by Vanderbilt University. The prosthesis provides powered knee flexion and extension through a range of motion from −5° (hyperextension) to 115° of flexion and powered ankle dorsiflexion and plantar flexion from 45° of plantarflexion to 25° dorsiflexion. Embedded prosthesis sensors measure knee and ankle joint angles, velocities, and motor currents, prosthesis load using a load cell, and prosthesis motion using a 6-degree of freedom inertial measurement unit, i.e., accelerometers and gyroscopes. The third generation powered knee and ankle prosthesis, with a custom carbon fiber footplate and standard foot shell, is ∼4.75 kg in weight (15). The training concepts have also been applied and further developed in this study, while training users to walk on the Open Source Robotic Leg (37) and the lightweight robotic knee prosthesis (38). These same concepts can be applied to other powered lower limb devices. Each prosthesis is controlled using a finite state machine controller, and each ambulation mode is divided into four phases: early to mid-stance, late stance, swing flexion, and swing extension. Each phase provides a different prosthesis response to mimic near normal kinematics of level-ground walking, incline walking, and stair ascent and descent. The sensors that detect prosthesis transitions throughout these phases include the

TABLE 1 | Participant demographics and predicate device description.

ID	Years post-amputation	Gender	Etiology	Age (yr)	Height (cm)	Weight (kg)	K-Level	Prescribed knee	Suspension type
TF01	0.75	M	Right sarcoma	28	193	73	K3	Ottobock C-Leg®	Skin fit suction
TF02	1	M	Left traumatic	32	188	81	K3	Ottobock C-Leg®/X3®	Suction with TES
TF03	1	M	Left traumatic	48	195	94	K3	Ossur Rheo Knee® XC	Seal-in liner
TF04	2	M	Left sarcoma	68	177	79	K3	Ottobock C-Leg®	Skin fit suction
TF05	2	M	Right sarcoma	33	177	63	K3	Ottobock Genium™	Seal-in liner
TF06	3	M	Right trauma	38	177	91	K3	Ottobock C-Leg®	Skin fit suction
TF07	4	M	Right-Infection	31	175	79	K3	Ottobock 3R80	Seal-in liner
KD08	5	M	Right traumatic	36	180	77	K4	Ossur Total Knee®	Liner and lock
TF09	7	F	Left sarcoma	26	160	52	K4	Ottobock C-Leg®	Skin fit suction
TF10	8	M	Right sarcoma	41	183	103	K3	Freedom Innovations Plie®	Liner with lanyard
KD011	9	M	Right sarcoma	26	177	91	K4	Ottobock Genium™	Seal-in liner
TF12	11	M	Right traumatic	19	185	62	K3	Ottobock Genium™	Seal-in liner
TF13	11	M	Left sarcoma	32	193	104	K3	Ottobock C-Leg®	Seal-in liner
TF14	14	M	Left traumatic	27	175	78	K3	Ottobock C-Leg®	Seal-in liner
TF15	15	M	Left traumatic	63	165	99	K3	Ottobock Genium™	Seal-in liner
TF16	15	F	Right sarcoma	29	170	70	K3	Ottobock C-Leg®	Skin fit suction
TF17	17	M	Right traumatic	55	168	64	K3	Ottobock C-Leg®	Suction
TF18	17	F	Right sarcoma	38	170	66	K3	Ossur Mauch®	Skin fit suction
KD19	18	M	Left sarcoma	33	187	86	K4	Endolite Hydraulic	Skin fit suction
TF20	18	M	Right traumatic	55	187	82	K3	Ottobock Genium™	Liner and pin lock
TF21	19	M	Left traumatic	47	182	97	K4	Ossur Total Knee®	Seal-in liner
TF22	20	M	Right sarcoma	29	170	60	K3	Ottobock 3R016	Liner with pin
TF23	24	F	Right traumatic	50	165	62	K4	Ossur Rheo Knee®	Sub-ischial vacuum
TF24	29	F	Left sarcoma	36	170	73	K3	Freedom Innovations Plie®	Liner with pin lock
TF25	32	F	Right infection	58	175	69	K3	Ottobock 3R60	Liner and pin lock
TF26	35	F	Right sarcoma	52	163	68	K3	Ottobock C-Leg®	Seal-in liner
TF27	38	M	Right traumatic	69	175	86	K3	Ottobock C-Leg®	Sub-ischial vacuum
TF28	39	M	Left traumatic	56	189	111	K3	Ottobock 3R80	Liner with TES belt
TF29	46	M	Left traumatic	61	180	84	K3	Ossur Mauch®	Skin fit suction
TF30	47	M	Left traumatic	50	190	106	K4	Ottobock 3R80	Liner with pin lock

individual's load and the prosthesis joint positions and velocities. The specific details of how the user can interact with the device and move throughout the phases of each mode are described in detail in subsequent sections.

Patient Training

Training begins by educating the user on the physical components of the device and the differences compared to their prescribed daily use prosthesis. The focus is to highlight the ability of the prosthesis to provide power in knee flexion and extension, and ankle plantarflexion and dorsiflexion across multiple ambulation modes. The majority of the K3/K4 level individuals, who fit the powered knee and ankle prosthesis in this paper, were able to independently traverse all modes available with the device within 3–6 h of instruction. The majority of the training session time is dedicated to adjusting prosthesis parameters to improve gait kinematics based on user and clinician feedback.

Users begin in the parallel bars with the prosthesis in standing mode. A gait belt should be used during initial safety training. In standing mode, they will be able to perform multi-directional weight shifting, including a single-limb stance, to gain confidence. Users will immediately notice the increased motion at their ankles compared to their passive device. The user is educated on the benefits of this available range of motion, including that it allows the prosthetic foot to remain flat and in contact with the ground during various ambulation modes (e.g., foot flat position during incline walking, allows the entire foot to be placed on a stair for ascending and descending steps, more comfortable sitting position and improved pre-positioning prior sit to stand transfers). For each mode, similar to the standard of care, the clinician will observe both swing and stance phases of the sound and prosthetic limbs in the frontal and sagittal planes, trunk position, and arm swing. Based on training information in this paper, clinical judgment is used to decipher between user causes for a particular gait deviation vs. a parameter adjustment to the device. Verbal and tactile instructions are given for improved symmetry, upright posture, and equal weight bearing to achieve desired outcomes before any prosthetic parameter changes. If the user is displaying any

FIGURE 1 | Level-ground walking with powered knee and ankle prosthesis.

TABLE 2 | Walking mode deviations, user instruction, and prosthesis parameter adjustments.

Mode	Deviation	User instruction	Prosthesis setting
Walk	Unable to initiate swing	Cue to increase sound side step length for increased stance time on prosthesis.	Decrease pre-set minimum dorsiflexion angle for ease of swing phase initiation
	Decreased foot clearance	Cue to stand tall and utilize hip flex during swing phase.	Increase knee flexion angle during swing phase for more clearance
	Excessive hip flexion, vaulting, hip hiking, and/or circumduction	Cue and/or use a mirror to improve awareness of foot position and to decrease excessive hip motions.	N/A
	Uneven heel rise	N/A	Increase or decrease knee flexion angle during swing phase to modify heel rise
	Insufficient swing speed	N/A	Increase swing extension knee stiffness to improve swing speed
	Rapid plantar flexion at heel strike	N/A	Increase early to mid- stance ankle stiffness and/or damping

deviations of the trunk, such as lateral bending, decreased arm swing, or decreased trunk/pelvis rotation, the clinician should assess the socket fit and comfort.

Level-Ground Walking

Goals

Clinical goals include the ability to ambulate (**Figure 1**) without upper extremity support, with equal step length, arm swing, and trunk rotation, and at near desired speed without limitations or noticeable gait deviations.

Prosthesis Control

While walking, as the user progresses forward in stance phase over the forefoot of the prosthesis, the powered ankle dorsiflexes. When the ankle dorsiflexes past a pre-set dorsiflexion angle (usually 6–8 degrees), the prosthesis will transition to the late stance phase and begin to provide powered plantarflexion. As the user's load shifts from the prosthetic foot and onto their sound foot, the prosthesis transitions to the swing phase. The knee flexes and the ankle dorsiflexes to provide clearance and, then, actively extends to prepare the prosthesis for heel strike. Once a load is detected in the prosthesis, it will transfer into the stance phase to provide a stable knee, promote weight acceptance, and allow forward progression through stance. This cycle continues to provide steady-state level-ground walking.

Training

Training of level-ground walking should begin in the parallel bars to allow for upper extremity support if needed. Gait assessment is completed for both stance and swing phase of walking, while appropriate modifications are made to the powered leg parameters (**Table 2**) (26). During initial training, it is beneficial to instruct the user to step with their sound limb first, while providing tactile and verbal cues to increase stance time on the prosthesis, which will assist with swing initiation. While standing behind or to the side of the user, tactile cues may include

physical assistance to provide force at the user's iliac crests to guide and hold their load onto the prosthesis. Additionally, verbal instruction is provided to encourage a larger sound sidestep, resulting in increased stance time on the prosthesis. The transition from late stance phase to swing phase will likely feel different from his/her prescribed device due to the ankle providing powered plantarflexion during late stance to assist with push off and swing. When necessary, a mirror can provide visual feedback to assist with prosthetic placement and to improve posture/trunk positioning.

If the prosthetic knee is extending too quickly during the swing phase and resulting in a forceful terminal impact, it is important to assess the user's interaction with the prosthesis. Many users are accustomed to providing a forceful hip flexion motion to advance their prostheses. This excessive motion is no longer needed since the device can provide powered swing extension. Verbal prompts to lessen hip flexion motion, by providing awareness to the user that the prosthesis is providing adequate swing clearance and the increased hip flexion movement is not necessary, may diminish the deviation. If the excessive terminal impact continues, the knee extension parameters are adjusted to reduce the speed of swing extension. Alternatively, if the leg is not extending quickly enough, he/she may be walking quicker than the device's initial settings allow. Swing extension parameters should be adjusted to decrease swing time and accommodate the user's speed.

Decreased clearance of excessive heel rise during the swing is corrected by adjusting the swing flexion parameters. Increased hip flexion during swing or vaulting of the sound limb could also be demonstrated by users. This may be due to habit or caused by being unaccustomed to the feel of the device and its active ankle/foot mechanism. Clinical reassurance that the user has appropriate clearance during swing due to the active dorsiflexion, and/or providing visual feedback with the use of a mirror may minimize this deviation.

Demonstration of forceful heel strike and strong hip extension is commonly observed with the prescribed passive device to ensure full knee extension and foot placement for initial contact. Similar to microprocessor or other stance control passive knees aligned to allow knee flexion, further education is provided that the knee does not need to be fully extended to accept their weight at heel strike; the leg will support them when a load is detected. Once this is addressed, ankle stiffness and/or knee extension parameters in early to mid-stance can be adjusted for comfort and allow a smooth weight acceptance.

The amount and timing of powered plantarflexion should be monitored to ensure it is comfortable for the user and does not interfere with foot clearance during swing. If necessary, the amount of powered plantarflexion can be reduced during initial training. Additionally, the user may show difficulty initiating the swing phase. This can be pronounced if the user displays a shortened step length with their intact limb and decreased stance time on the prosthesis, as often seen with passive devices. Verbal and tactile cues through palpation and contact guard assist with the gait belt to guide the user to take longer sound sidesteps and increase weight bearing through stance to allow the prosthesis to swing with more natural timing. The pre-set dorsiflexion ankle angle, nominally set to 6–8 degrees of dorsiflexion, can be reduced to ease swing initiation when feedback to the user is not effective. This necessary dorsiflexion angle is what allows the prosthesis to transition to the late stance phase and for the leg to begin to provide powered plantarflexion. While training an amputee, who may be hesitant or tends to have a step-to gait with their prescribed prosthesis, decreasing the dorsiflexion angle parameter may allow ease of transitioning into swing during training. Once the user becomes more comfortable with the device and begins to show increased step length, this parameter is often adjusted back to the starting range. Once the user is walking comfortably within the parallel bars, the walking distance can be increased, and the user should be able to ambulate with decreased upper extremity support. Individuals often adapt quickly and achieve improved swing initiation in a longer walkway as they gain confidence in stance and demonstrate their ability to increase their stance time on the device. This feature of stance stability can promote increased step length and stance time, while also improving trunk/pelvic rotation and arm swing. Usually within 5–10 min of level-ground walking training, the K3/K4 level users can walk comfortably without assistance or significant gait deviations.

Stair Climbing

Goals

Clinical goals include the ability to ascend and descend stairs (**Figure 2**) with unilateral upper extremity support, achieve reciprocal stepping without cueing, demonstrate consistent foot placement to achieve appropriate power initiation, and demonstrate controlled lowering during reciprocal stair descent.

Prosthesis Control

During stair ascent (26, 39), as the weight of the individual shift off the powered prosthesis, the device transitions to the swing phase, where the knee flexes (∼90 degrees) to provide proper stair clearance. The knee, then, extends slightly and the ankle dorsiflexes (5–10 degrees) to prepare the foot for placement onto the next step. As the user shifts his/her weight onto the prosthesis, the device transitions to the stance phase, where the knee provides powered extension and the ankle provides stability as it moves under load toward the neutral position. Once the knee is fully extended, the user can position their sound limb on the next step. As they unload the prosthesis, it can provide powered plantarflexion, followed by powered knee flexion to provide clearance, and prepare for the next step.

During stair descent, as the user shifts their weight onto the prosthetic foot, the powered knee and ankle provide resistance to support the user as they "ride" the knee down for controlled descent into knee flexion and ankle dorsiflexion. As the user continues to progress through a stance of stair descent and begins to shift their load onto their sound limb, the powered prosthesis will activate the swing phase of stair descent. This will allow knee flexion and ankle dorsiflexion to clear the step and reposition for the next step.

FIGURE 2 | Demonstration of stair climbing with powered knee and ankle prosthesis.

TABLE 3 | Stair climbing deviations, user instructions, and prosthesis parameter adjustments.

Mode	Deviation	User instruction	Prosthesis setting
Stair ascent	Poor foot placement	Cue for proper body and foot position to prepare for stair ascent	Adjust swing extension phase ankle dorsiflexion angle to achieve a foot flat position
	Decreased foot clearance	Cue to stand tall and utilize hip flex during swing phase	Increase swing phase knee flexion angle for more clearance
	Vaulting, hip hiking, circumduction	Verbal and tactile cues to improve awareness of foot position and to limit excessive hip motions.	N/A
	Inadequate support and power during stance	Cue to increase stance time on prosthesis and decrease upper extremity support	Increase stance phase knee stiffness for improved support and power into knee extension
	Insufficient swing speed	N/A	Decrease or increase rate of swing flexion
Stair descent	Unable to initiate knee flexion at initial contact	Cue for body and foot position to prepare for stair descent	Decrease stance phase knee damping to allow for easier knee flexion at initial contact
	Inadequate support during stance	Cue to increase stance time on prosthesis	Increase stance phase knee damping for increased support into knee extension during early stance phase Increase stance phase knee stiffness for increased support into knee extension during mid to late stance phase
	Poor foot placement	Cue for proper body and foot position to prepare for stair descent	N/A
	Decreased foot clearance during swing	Verbal and tactile cues to increase stance time on prosthesis and decrease UE support	Increase swing phase ankle damping

Training

Reciprocal stair climbing should begin on a staircase with four or fewer stairs and bilateral handrails. Instruction begins with verbally describing the motions of the powered prosthesis during reciprocal stair ascent since most users of mechanically passive devices utilize a step to pattern of stair climbing using their sound limb to raise them to each step. Since the powered ankle can provide active dorsiflexion, users can place their whole foot onto the step and achieve a flat foot position during both stair ascent and descent, which can allow for a greater sense of stability during reciprocal stair climbing.

Training begins by ascending a single step to prepare the user for the movement and feeling of powered knee extension. While standing in front of the stair, the user shifts their weight off the prosthesis to allow the powered knee to transition into swing flexion. The user is, then, instructed to perform active hip flexion to raise the prosthesis and place the prosthetic foot fully onto the step. Physical cueing with hand placement at the

FIGURE 3 | Incline walking with powered knee and ankle prosthesis.

TABLE 4 | Ramp mode deviations, user instruction, and prosthesis parameter adjustments.

Mode	Deviation	User instruction	Prosthesis setting
Ramp ascent	Decreased foot clearance	Cue to stand tall and utilize hip flex during swing phase	Increase knee flexion angle during swing phase for more clearance
	Vaulting Hip hiking Circumduction	Cue and/or use a mirror to improve awareness of foot position and to decrease excessive hip motions	N/A
Ramp descent	Unable to initiate knee flexion at initial contact	Cue to increase stance time on prosthesis and decrease upper extremity support	Decrease stance phase knee damping to allow for easier knee flexion at initial contact
	Inadequate support during stance	N/A	Increase stance phase knee damping for improved support into knee extension during early stance phase Increase stance phase knee stiffness for improved support into knee extension during mid to late stance phase

lateral hip to guide the prosthetic side to assist with hip flexion and resist circumduction and vaulting. The user is instructed to push down into the prosthesis, producing pressure toward the distal/posterior portion of the socket and creating a hip extension moment. This transfer of weight onto the prosthesis is detected by the load sensor and activates the prosthetic knee extension power. Verbal instructions are given to encourage a slight forward trunk lean to assist with balance. Once the full-powered knee extension is achieved, the user will place their sound limb next to the prosthesis on the first step. Several trials of ascending one step are performed until the user feels comfortable with the movement. Stance phase stiffness parameters swing phase knee and ankle clearance, and foot position can be adjusted based on user and clinician preferences (**Table 3**).

Once the user can ascend one step comfortably, he or she can progress to climbing up several steps in a reciprocal pattern, starting with their sound limb. They are reminded that shifting weight off the prosthesis will cause the knee to swing and prepare for prosthetic foot placement onto the next step. While standing behind the user, a contact guard assist with the use of the gait belt or physical palpation at the user's hips is provided and should continue to be provided to assist the users with weight-shifting and loading of the prosthesis, body position, upper extremity support, and proper foot placement. The clinician should continue to monitor swing phase clearance, quality of knee extension, foot placement, and adjust prosthesis parameters as appropriate. The amount of desired knee extension power may change throughout training as the user begins to increase their weight-bearing through the device and decrease their reliance on upper extremity support; stance phase knee stiffness can be increased to provide more support. One goal is to have users progress to only using the handrails for balance assistance (preferably only one handrail), and cues can be given to prevent the user from lifting or pulling up the step. This may take several trials to determine the appropriate power level and for the user to

gain confidence in the ability of the prosthesis. If a circumduction or a sound side vaulting occurs, swing phase parameters can be adjusted to confirm stair clearance. The user's socket comfort should be monitored and modified as needed due to increased hip flexion of the amputated limb during stair ascent.

During stair descent, the method is similar to riding the knee down with a passive prosthesis, but users are reminded that for the powered prosthesis, the whole foot can remain on the step for added stability. Instructions are given to start with the prosthesis side first during descent and load the prosthesis to ride the knee down. Controlled knee flexion is achieved with stance phase stiffness and damping parameters (**Table 3**). When their sound side reaches the next step and they shift their weight off the prosthesis, it will swing toward knee extension in preparation for the following step allowing them to continue in a reciprocal pattern. Physical support is provided while standing behind the user with assistance at the gait belt for weight shifting onto the prosthesis and palpation at the user's shoulder to promote upright posture. Verbal instruction will be provided for foot placement on the step, and hand placement on the railing for balance stability. Clinical observations of foot placement, controlled lowering, and swing clearance should be made and prosthesis parameters can be adjusted as needed. Feedback from the user is also needed to confirm comfort and ease of stair descent.

Incline Walking

Goals

Clinical goals include the ability to ascend and descend inclines of up to 10 degrees (**Figure 3**) with near equal step length, arm swing, and trunk rotation while using unilateral or no upper extremity support. Additionally, users should be able to demonstrate ramp descent with controlled lowering.

Prothesis Control

Control for ramp ascent mode (26, 28) is similar to level-ground walking mode. As the user loads the prosthesis through mid-stance the ankle will dorsiflex. When the ankle dorsiflexes past a pre-set angle (usually 8–10 degrees), the prosthesis will transition to the late stance phase and provide powered plantarflexion to assist with forwarding propulsion up the incline. Once the user shifts weight onto their sound limb, the powered knee flexion followed by powered knee extension will occur. Any parameter changes for level-ground walking should be transferred and used as the starting point for initial ramp ascent training.

During ramp descent, as the user loads the prosthesis, the powered knee and ankle provide resistance to support the user as they "ride" the knee down for controlled descent into knee flexion and ankle dorsiflexion. As the user progresses through stance and the ankle dorsiflexes past a pre-set angle, the prosthesis will progress through terminal stance. The swing will occur when decreasing prosthesis load is detected, as the user transfers their weight to their sound side.

Training

Training of incline walking begins on a slope with bilateral handrails. The user is instructed to ambulate up the ramp with bilateral upper extremity support, even step length, and a slightly forward posture to assist with propulsion up the incline. Many of the deviations seen and resolved during level-ground walking can be addressed in similar ways during incline walking (**Table 4**). Verbal reminders of how powered plantarflexion and powered swing extension can assist users up the ramp are beneficial, since the technology is different from their prescribed prosthesis. These motions are more pronounced during ramp ascent than in level-ground walking. During incline walking, users will likely have a greater awareness of the ankle's available range of motion into dorsiflexion, which allows the foot to remain flat on the incline during early to mid-stance.

During ramp descent, the user is instructed to take shorter steps during initial training to assist with weight-bearing onto the device and to "ride" the knee into flexion. If individuals are not currently using their prescribed device's stance resistance for ramp descent, this training may require several trials for them to feel comfortable putting weight through the device as the knee bends and trusting the resistance during stance. User feedback and clinician expertise are used to select parameters (adjusting knee stiffness and damping) to remove the feeling of the user "falling" down the ramp and diminish the impact on the sound limb. The clinician will observe upper extremity support and provide additional cues as the user becomes more comfortable with the powered knee stability and increase weight bearing through the prosthesis. Additional physical cues at the shoulder and hip to guide the user onto the prosthesis and direct their load down through the device to verify needed assistance for the user to adequately descend the ramp at their desired speed and support. The individual should ambulate up and down the ramp as needed while receiving cues from the clinician and parameter adjustments to achieve the clinical goals stated above.

Sit to and From Standing

Goals

Clinical goals include the ability to rise from a seated position (**Figure 4**), with or without upper extremity support, demonstrate consistent foot placement and trunk position to achieve appropriate power initiation and comfortable standing without cueing, and demonstrate controlled lowering when completing standing to seated movements.

Prothesis Control

Sitting transfers are divided into four phases: stand-to-sit, relaxed sitting, sit-to-stand, and standing (29, 40). The stand-to-sit phase occurs when the user loads the prosthesis and creates a sustained knee flexion moment above a pre-set threshold. Damping and stiffness parameters in this phase will allow the user to have controlled resistance into knee flexion and ankle dorsiflexion to lower themselves to a chair. Once seated, as defined by the knee and ankle joints crossing, a pre-set flexed threshold and joint velocities are close to zero, the prosthesis transitions to the relaxed sitting phase. In the relaxed sitting phase, the knee and ankle joint remain compliant and can be easily repositioned manually by the user to a comfortable position. To initiate the sit-to-stand phase, the user shifts his/her weight onto the powered prosthesis. As load increases over a pre-set threshold,

FIGURE 4 | Sit-to-stand and stand-to-sit weight transfers with the powered knee and ankle prosthesis.

TABLE 5 | Sit-to-stand and stand-to-sit deviations, user instructions, and prosthesis parameter adjustments.

Mode	Deviation	User instruction	Prosthesis setting
Stand to sit	Unable to initiate sitting	Cue for active hip flexion and equal weight bearing between limbs for increased load onto the prosthesis	Decrease axial load threshold
	Inadequate support during sitting	N/A	Increase or decrease knee damping to provide more or less support, respectively
Sit to stand	Unable to initiate standing	Cue to increase forward trunk position and load onto the prosthesis	Decrease axial load threshold to initiate knee extension power
	Inadequate support or power during standing	Cue for equal weight bearing between limbs for increased load onto the prosthesis	Increase stance phase knee stiffness for improved support and power into knee extension

the device provides powered knee extension and powered ankle plantarflexion (from a dorsiflexed position to a neutral angle) to aid the user in rising to a standing position. Once the user is standing upright with full knee extension, the prosthesis transitions to the standing phase.

Training

Begin sit-to-stand training with the user standing within parallel bars with a chair of standard height, with armrests positioned closely behind them. The powered knee and ankle prosthesis offer a controlled descent by providing support throughout the full knee flexion and ankle dorsiflexion range of motion to a seated position. While standing, they are instructed to have equal weight through each leg and apply a knee flexion moment by attempting to sit with active hip flexion and a forward trunk lean. It is beneficial to provide verbal instruction on trunk position and physical assistance through the user's hips and spine to encourage equal weight bearing through the lower extremities while the prosthesis flexes to a seated position. The user may utilize the armrests for balance if needed while transferring to the chair. The clinician should monitor foot and trunk position, loading through the device, and rate of controlled flexion and adjust parameters as needed (**Table 5**). For sitting down, the knee and ankle joint damping parameters can be increased or decreased for more or less support, respectively. As the user becomes comfortable with the movement into sitting, they may start to increase their load onto the prosthesis and parameters can be further adjusted for increased stiffness.

Once the user is seated and in the relaxed sitting phase, they may adjust the prosthesis passively to their desired position for sitting or to prepare to stand. The available range of motion at the knee and ankle joint allows the user to scoot toward the edge of the chair, align their feet evenly for the equal load on both limbs and maintain a flat foot position. This position will enable bilateral limb muscle activation and improved pelvic symmetry for a smoother, more efficient transition to standing. For users who have difficulty initiating load in the prosthesis to facilitate powered knee extension, enabling visual feedback of the amount of load in their prosthesis allows both users and clinicians to become accustomed to the amount of forward lean and load that is needed to initiate stand without engaging power at the device may be helpful.

Clinicians should evaluate the ease of initiating standing along with the rate and movement quality of rising to stand, while prosthesis parameters can be adjusted as needed (**Table 5**). Tactile cues are provided to the user along their torso to encourage forward lean, and with the gait belt to pull the user toward their prosthetic side to increase weight-bearing through the prosthesis while standing up. Verbal cues and demonstration of proper foot positioning and posture will assist for successful sit-to-stand transitions. If the user is having difficulty initiating sit to stand, and cues don't resolve the issue the axial load threshold can be decreased for an easier transition. Stance phase knee stiffness can be increased for increased support and power into knee extension to achieve standing. If there continues to be user hesitation to load the prosthetic foot from sitting to standing, training techniques may include staggering their feet; placing the prosthetic foot slightly behind the sound foot will force an increased weight bearing on the prosthesis. An alternative method is to provide support in front of the user and guide their upper extremities and trunk forward and slightly toward their prosthetic side. By being present in front of the user, they feel more secure and may allow themselves to lean forward over their toes to increase load through the device.

RESULTS

All 30 participants were successful in powered leg-fitting and ambulation over level-ground (see **Supplementary Video 1**). Twenty-seven participants were able to continue with training sessions and became successful in independent ambulation of all other functional modes after participating in 1–3 sessions of 1–2 h in length. Three of the participants did not continue with additional powered leg ambulation training. Two subjects (TF02 and TF15) were unable to continue with the training of stair climbing and ramp ascent due to fatigue and tolerance of the weight of the powered prosthesis required for stair ascent and incline walking. Another subject (TF17) was unable to continue due to the cognitive task of reciprocal stair climbing; he required multiple cues for foot placement and a residual limb control required for stair climbing.

Several of the subjects were trained on multiple powered legs throughout the development: 27 of the participants were trained on the Vanderbilt powered knee-ankle prosthesis, 14 of the participants were trained on the OSL, and 11 of the participants were trained on the lightweight robotic knee with a passive ankle [low-profile Vari-Flex foot (41)]. Nine of the participants had the opportunity to train on all 3 powered leg prostheses. **Figure 5** outlines averaged prosthetic leg knee and ankle kinematics across all trained ambulation modes.

DISCUSSION

The current market for transfemoral amputees and the passive prosthesis is very focused on the functional level of the amputee to which device they would be best paired with for success. The functional level is decided by the clinical team based on the amputee's current and potential functional status. Many of the participants enrolled in our study are unable to climb a staircase reciprocally or descend an incline forward with the use of their currently prescribed prosthesis but were successful using a powered prosthesis. Powered knee and ankle prostheses have the potentially to truly improve the amputee's activities of daily living based on proper training and device control and development.

Over the past 10 years of developing these training techniques, participant feedbacks included that the powered prosthesis felt very different from their currently prescribed prosthesis, was more intuitive to use, allowed them to "walk without thinking at each step," and did not have to actively or forcefully move their hip forward to advance the prosthesis. Once trained, users were able to ambulate very comfortably as evident by holding conversations, carrying items, and navigating in tight spaces without noticeably increased effort. Following the development of this training, the majority of users, who are now being trained, are successful in ambulating across all five activity modes within 1–3 sessions of 1–2 h each. Occasionally a few participants needed 1–2 additional sessions to address socket fit or socket suspension issues due to the weight of the powered leg being greater than their prescribed prosthesis, and/or becoming more comfortable with the power and movements of the device.

Ambulation training was based on allowing the user to walk as they did before their amputation. Learning to climb stairs with a reciprocal pattern and/or stand up from a chair incorporating their residual limb and prosthesis was at first both a physical change (e.g., the prosthetic side would lead on every other step during stair climbing) and a cognitive change (e.g., users had to remember to engage their prosthetic side while standing up from a chair). While only higher level and very active users can demonstrate reciprocal stair climbing with a passive knee unit, since it requires extensive residual limb strength and stability, all participants in this study who tried the powered prosthesis were successful climbing with a reciprocal gait. A few participants required additional cueing for stair climbing for appropriate loading of the prosthesis and trunk position. Often this was during initial stair ascent training and after 2–3 successful steps, they begin to have more trust in the movements, increase their load through their socket, lessen their upper extremity support, and relax their trunk position into a more natural posture. Participants often expressed excitement to have the ability to ascend several stairs and even staircases with ease due to the active power provided by the prosthesis. The users responded positively when given the ability to rely on the prosthesis when rising from a chair and reported decreasing load on their sound limb and support on the arms of the chair. Positive reactions were also expressed when walking up a large incline since the effect of the ankle power when ascending a ramp was felt immediately. Several individuals stated they were able to walk up the incline faster, with improved ease, and with less (or no) reliance on upper extremity support. Anecdotally, it was observed for some users that sound side vaulting was minimized or diminished during level-ground and incline walking with the powered knee and ankle prosthesis without any specific instructions or cueing when walking.

Only a small percentage of participants (3 of 30) were unable to complete full training of all ambulation modes with a powered

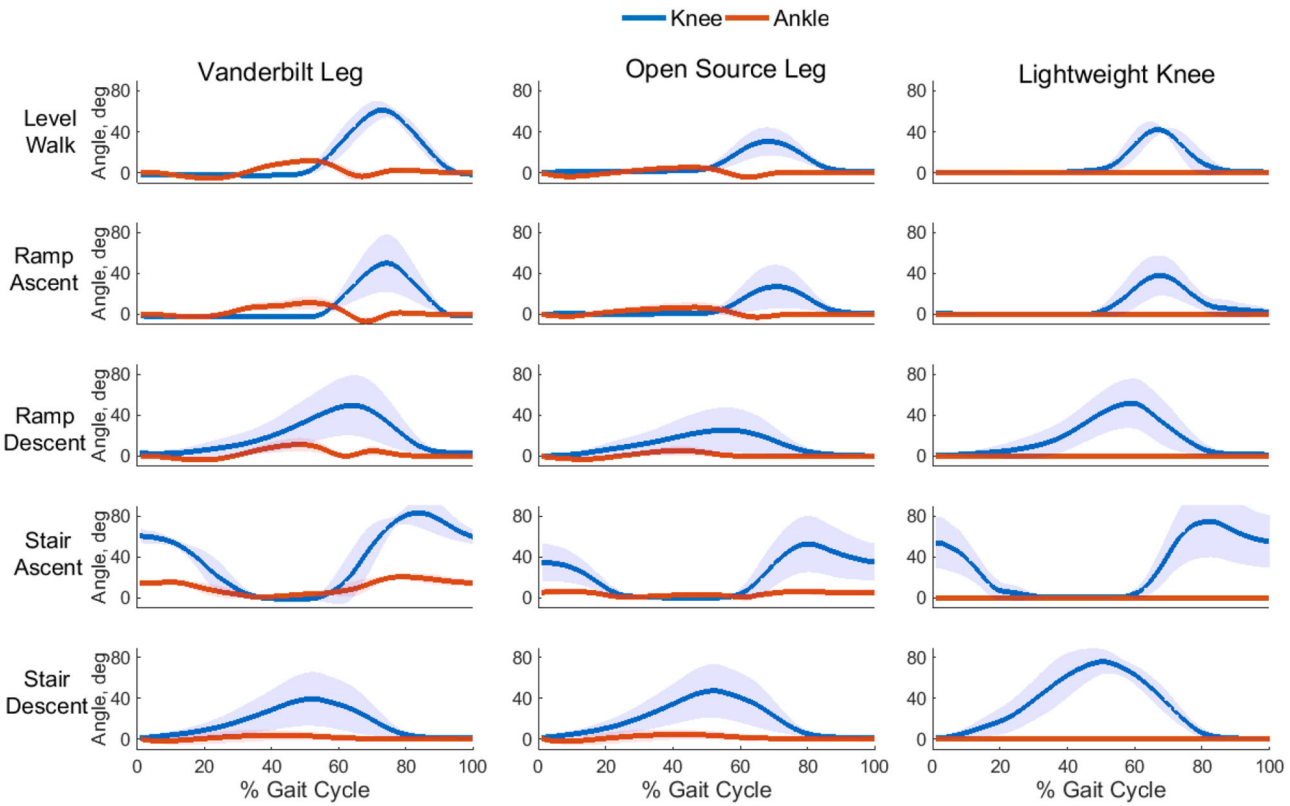

FIGURE 5 | Averaged powered prosthesis knee and ankle angles during all ambulation modes for all participants. Shaded regions represent +/– one standard deviation (SD).

leg prosthesis. Two individuals were independent with level-ground ambulation, but experienced difficulty when needing to lift the prosthesis either upward for stair ascent or forward up an incline. The passive lifting required by the user's hip and abdominal muscles was difficult for these two participants who had short residual limbs, resulting in an increased load onto their limb and hip musculature. Additionally, one individual was unable to perform the cognitive task of reciprocal stair climbing. Although he may have been able to eventually re-learn this task, additional sessions in the research environment were not available at the time. While the majority of subjects provided positive feedback while using the powered knee and ankle prosthesis, several reported the desire for the leg to be lighter, quieter, and have water-resistant capabilities to allow them to use the device with all desired activities.

The kinematic data shown in **Figure 5** were based on user and device testing across several years of research on three different powered leg prostheses, multiple users on each device, and at different times during powered leg training development. These data in this paper intend to demonstrate kinematics of successful use across multiple modes of ambulation and not necessarily to compare between devices. Most of the differences, if not all, identified in **Figure 5** can be explained by the differences in hardware and/or improvements in control that developed over time. Additionally, the testing of these devices was rather sequential: testing of the Vanderbilt Leg spanned from 2011 to 2018, Open Source Leg from 2017 to 2021, and the Lightweight Knee from 2019 to 2021. For example, the Vanderbilt Leg prosthesis had 70 degrees range of motion available (15) at the ankle, whereas the Open Source Leg only had 30 degrees range of motion available (42). Therefore, ambulation on the Vanderbilt Leg, compared to the Open Source Leg, could take advantage of this increased range of motion including increased stance phase dorsiflexion during stair and ramp descent and late-stance powered plantarflexion during level-ground and incline walking. Additionally, as we became more proficient in our control settings for powered leg prostheses, we realized that for adequate toe clearance, we did not need to flex the knee as much during the swing phase (e.g., in **Figure 5**, comparing maximum knee flexion during the swing phase of walking with the Vanderbilt Leg to that of both the Open Source Leg and Lightweight Knee). Had we identified this improvement earlier in our development, we could have easily adjusted with the Vanderbilt Leg to result in similar knee kinematics between the legs and hence, similar swing clearance for the users.

These data did, however, help in developing this training protocol of how to teach individuals with a transfemoral amputation and how to walk on a powered leg prosthesis.

Simultaneously, as we were learning to control a powered prosthesis, we were developing the appropriate clinical training cues based on the feedback received. The speed at which our users became accustomed to the device was quicker than initially expected (e.g., we progressed training to inclines and stairs much sooner than anticipated).

We now have a training protocol for powered lower limb prostheses that we hope will assist other research groups, including our own, in providing training for these devices before performing studies that involve functional performance outcome measures and/or biomechanics. These comparative studies will be important to assist in identifying when (e.g., which ambulation tasks) and where (e.g., ankle only, knee only, or both knee and ankle) users can best take advantage of the power available from these devices. Although the data included in this study cannot make these comparisons, we were surprisingly successful in training users from the various demographic backgrounds; participants with a wide range of time since amputation (9 months−47 years), height (160–193 cm), weight (52–111 kg) all had similar training time. A female who was 160 cm and 52 kg and 7 years post-amputation completed all modes as easily as a male who was 186 cm, 111 kg, and 39 years post-amputation. Both users were able to ascend stairs and ramps with the same instruction and ease with initial parameter and joint power settings based on their weight.

Ambulation and negotiation of the various activity modes were also successful across individuals with a variety of suspension systems, provided that the setup used could accommodate the increased weight of the powered device. Participants that used a pin-locking liner did display increased rotation during the swing, likely due to the active knee power. Since we did not change individuals' primary method of suspension, the addition of a Total Elastic Suspension belt for these users eliminated the rotation. A TES belt for secondary suspension was also necessary for users who presented with shorter residual limbs. Users with shorter residual limbs often required supplementary training to properly lift and load the powered prosthesis. Palpation at the user's hips and lower back to provide tactile cues to incorporate hip flexors and abdominal muscles, decrease posterior lean aided for proper prosthetic side foot placement on a stair to properly load their socket using active hip extension when ascending stairs, and complete sit to stand transitions allowed for the prosthesis to respond with active knee extension.

While these training methods were developed using three different powered leg prostheses [i.e., Vanderbilt Powered Knee and Ankle Prosthesis (15), the Open Source Robotic Leg (37), and the lightweight hybrid robotic knee (38)] in a rehabilitation facility environment, we expect most of the methods to transfer to similar powered lower limb devices. Additional training techniques may be necessary for outdoor/uneven terrain ambulation and obstacle avoidance for participants to function independently in their home environment. These methods were developed while training high-level (K3 and K4) ambulators with non-vascular reasons for amputation. The duration or frequency of training may change for K2 ambulators. Additional cues may be needed to load the prosthesis during ascent activities secondary to decreased strength or balance deficits or to incorporate an assistive device. Finally, since training occurred on a prosthesis that is not yet clinically/commercially available, all participants attended training sessions and, then, returned to their prescribed and passive prosthesis at the end of the research sessions.

CONCLUSIONS

As powered lower limb devices become more clinically available, they will continue to challenge physical therapy practice in terms of instructional gait and advanced mobility training. Through this training protocol, clinicians can gain a better understanding of the technical aspects of how the device is controlled, as well as the benefits and limitations, to provide better training and outcomes for users of lower limb prostheses across multiple modes of ambulation. Physical therapists should be encouraged to study and understand these devices through education from prosthetists, manufacturers, and published research studies and protocols.

DATA AVAILABILITY STATEMENT

The raw data supporting the conclusions of this article will be made available by the authors, without undue reservation.

ETHICS STATEMENT

The studies involving human participants were reviewed and approved by Northwestern University Institutional Review Board. The patients/participants provided their written informed consent to participate in this study. Written informed consent was obtained from the individual(s) for the publication of any potentially identifiable images or data included in this article.

AUTHOR CONTRIBUTIONS

LH conceived of the original idea and supervised the project. SF and AS performed the subject training, data collection sessions, and developed the subject training process. AS and LH analyzed the data. SF, AS, and LH wrote and edited the manuscript. All authors contributed to the article and approved the submitted version.

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SUPPLEMENTARY MATERIAL

The Supplementary Material for this article can be found online at: https://www.frontiersin.org/articles/10.3389/fresc.2022.790538/full#supplementary-material

Supplementary Video 1 | Clinical training goals and example videos of individuals learning to use a powered prosthesis for level and incline walking, stair climbing, and during weight transfers.

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BRIEF RESEARCH REPORT
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A Haptic Sleeve as a Method of Mechanotactile Feedback Restoration for Myoelectric Hand Prosthesis Users

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Current myoelectric upper limb prostheses do not restore sensory feedback, impairing fine motor control. Mechanotactile feedback restoration with a haptic sleeve may rectify this problem. This randomised crossover within-participant controlled study aimed to assess a prototype haptic sleeve's effect on routine grasping tasks performed by eight able-bodied participants. Each participant completed 15 repetitions of the three tasks: Task 1—normal grasp, Task 2—strong grasp and Task 3—weak grasp, using visual, haptic, or combined feedback All data were collected in April 2021 in the Scottish Microelectronics Centre, Edinburgh, UK. Combined feedback correlated with significantly higher grasp success rates compared to the vision alone in Task 1 ($p < 0.0001$), Task 2 ($p = 0.0057$), and Task 3 ($p = 0.0170$). Similarly, haptic feedback was associated with significantly higher grasp success rates compared to vision in Task 1 ($p < 0.0001$) and Task 2 ($p = 0.0015$). Combined feedback correlated with significantly lower energy expenditure compared to visual feedback in Task 1 ($p < 0.0001$) and Task 3 ($p = 0.0003$). Likewise, haptic feedback was associated with significantly lower energy expenditure compared to the visual feedback in Task 1 ($p < 0.0001$), Task 2 ($p < 0.0001$), and Task 3 ($p < 0.0001$). These results suggest that mechanotactile feedback provided by the haptic sleeve effectively augments grasping and reduces its energy expenditure.

Keywords: haptic, mechanotactile, sensory feedback, sensory restoration, prosthetic, prosthesis, hand, upper limb

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INTRODUCTION

According to the closed loop theory of motor control, movement of a healthy human hand is governed by co-dependant feedforward muscle control and sensory feedback (1). Based on the latter, feedforward muscle control is adjusted to achieve economy of movement and the lowest possible metabolic energy expenditure (2), therefore closing the loop. When limb loss occurs, the loop of motor control becomes disrupted. The feedforward component of the loop may be partially restored with myoelectric prostheses (3). However, these devices do not restore sensory feedback,

leaving the loop of motor control open (4). As a result, prosthetic users only have uncertain feedforward control at their disposal (5), making them unable to perceive tactile properties of handled objects and experience diminished motor control (6). They cannot feel their prosthetic grip force, leading to application of excessive force (resulting in excessive energy expenditure and muscle fatigue) and crushing of handled objects (7–9). In order to achieve satisfactory prosthetic performance, users heavily rely on visual feedback, which in turn increases cognitive load (10, 11). For these reasons, prosthesis embodiment remains poor, as reflected in the prosthesis abandonment rate of 40% (12).

Considering the above, it is unsurprising that most amputees agree restoring sensory feedback is as important as restoring feedforward muscle control (7, 13–15). Restoring tactile feedback, a type of sensory feedback, is especially promising. It has been shown to not only significantly improve grasp success rate (16–18) but also significantly decrease grip force (19–22). Restoring tactile feedback is also predicted to reduce prosthesis abandonment rate (23), providing a strong rationale for development of tactile feedback modalities.

So far, invasive and non-invasive tactile feedback modalities have been developed (24). Invasive modalities, such as targeted sensory re-innervation, direct peripheral nervous system stimulation and central nervous system stimulation, are promising due to their potential to elicit near-natural touch sensations (25–27). However, their clinical utility remains challenging (28). They carry a number of risks, such as nerve damage (29) and paraesthesia (30), have been tested on a limited number of volunteers and face much reluctance from amputees (31).

Non-invasive modalities, such as vibrotactile, electrotactile and mechanotactile feedback systems (28) are comparatively better characterised and constitute a more acceptable alternative as they require no surgical interventions (31). Yet, non-invasive modalities are not without their caveats. The main criticism of vibrotactile and electrotactile feedbacks is centred around their dissimilarity to endogenous tactile feedback, making them difficult to understand. Both are discontinuous (composed of discrete vibration or electric current bursts) and modality mismatched (vibrations or electric currents felt on the skin usually encode grip pressure), contrary to biological feedback (32). In contrast, mechanotactile feedback is both continuous and modality matched (pressure applied to skin encodes grip pressure). As such, it mimics natural tactile feedback, making the artificial feedback more intuitive to understand (32, 33). However, these advantages are balanced out by mechanotactile devices being larger, heavier, and of greater energy demands than their electrotactile and vibrotactile counterparts, inhibiting their development (34).

Disadvantages associated with invasive and non-invasive tactile feedback modalities contribute to their clinical and commercial unavailability. Mechanotactile feedback, as the only non-invasive modality providing continuous and modality matched feedback, seems to have an underdeveloped potential. Hence, research into how its current caveats can be resolved is warranted.

This study aimed to test the utility of a new mechanotactile feedback restoration device, a prototype haptic sleeve. Haptic sleeves are sleeve-shaped, variable compression devices which have so far demonstrated utility in robot-assisted surgery (35), social touch mimicking (36), and virtual reality enhancement (37). They are lightweight and thin, addressing the problems of heaviness and large size characteristic of contemporary mechanotactile feedback devices. While haptic band devices have been developed to provide sensory feedback in rehabilitation robotics, they have not uniformly been integrated into a prosthetic sleeve which is an integral part of the socket (38–41). Where a pneumatic device has been integrated into the socket, it has been at a discrete point instead of providing distributed sensory feedback across the residual limb (42). Our study demonstrates a soft socket that integrates haptic feedback across its inner surface whilst being capable of supporting the terminal device without need for any additional material.

The primary aim of the study was to be achieved by assessing the impact of the haptic sleeve on grasp success rate and energy expenditure of grasping. Grasp success rate was chosen as it is a simple, concrete metric which is in wide use in tactile feedback restoration studies (16–18). However, it is an indirect measure of tactile feedback impact on motor control, making it difficult to speculate about a cause-and-effect relationship. Therefore, changes in energy expenditure of grasping were recorded, too, as they are a more direct and robust basis for establishing a causal link between feedback restoration and improvement in outcomes (2). It was hypothesised that using the haptic sleeve will result in higher grasp success rate and reduced energy expenditure of grasping.

MATERIALS AND METHODS

Design of the Haptic Sleeve

Wearable Sleeve

The prototype haptic sleeve used in the experiments was designed by the research team and manufactured by Koalaa Prostheses (London, UK) and can be seen in **Supplementary Figure 1**. Once mounted, the sleeve extended from the participant's proximal forearm to their wrist, allowing for a secure and comfortable fit while leaving enough space for the EMG electrode placement immediately distal to the elbow joint. The device was designed to compress the forearm proportionally to the pressure detected at prosthetic fingertips, thus delivering continuous mechanotactile feedback. To execute this function, the sleeve had a small motor (RS Pro 951D, RS Components, London, UK) mounted on its lateral side, as well as a pulley system with a thread wrapped around the sleeve equidistantly several times. Clockwise rotation of the motor resulted in winding of the thread around the sleeve, tightening it and therefore compressing the user's forearm. Anticlockwise rotation of the motor unwound the strings, untightening the sleeve and reducing the compression.

Electrical Design

To provide all the required analogue and digital inputs and outputs to the system, an Arduino Uno REV3 microprocessor (Arduino, Massachusetts, USA) was used. Connected to the

microprocessor via a protoboard were two force-sensing resistors (FSRs) (Interlink Electronics FSR400, California, USA), three separate push button switches, a L298N 2A motor driver (HandsOn Tech, Johor, Malaysia), adjustable power supply (30V 3A Tenma 72-2540) and a Y-bridge. The electronic circuit obtained functioned to detect force applied at FSRs on prosthetic fingertips and translate it into rotation of the haptic sleeve's motor. It also recorded the EMG signals whilst they were used for the myoelectric prosthesis control. Simultaneous EMG signal recording and use was enabled by the Y-bridge which split the EMG signals from the electrodes into two channels. One channel connected to the prosthesis, facilitating myoelectric control, while the other channel connected to the microprocessor, allowing signal recording.

Software/Hardware Interface

The microprocessor was programmed using the Arduino Integrated Development Environment (Arduino, Massachusetts, USA). Its main functions were to record and save experimental readings, as well as interpret the FSR readings by comparing their current averaged force readings to their previous averaged value. If the new value was greater than the previous one, the motor engaged for 0.1 s at the speed proportional to the new value, tightening the sleeve. However, if the new average force was smaller than the previous one, the motor retained its current position. In this way, the haptic sleeve could provide continuous, proportional mechanotactile feedback. The Pulse Width Modulation (PWM) of the motor at time t is defined by the following equations:

$$\overline{F}_t = \left(\frac{F_{Index} + F_{Thumb}}{2}\right) \quad (1)$$

$$PWM = \begin{cases} 0, & F_t \leq F_{t-1} \\ \frac{\overline{F}_t}{F_{MAX}} \times 100 \times PWM_{Max}, & F_t > F_{t-1} \end{cases} \quad (2)$$

Where $\overline{F}_t$ is the new average force, F_{t-1} equals the previous average force, F_{Index} is equal to the force from the index finger sensor, F_{Thumb} is the force from the thumb sensor, F_{MAX} equals the maximum force of the sensor and PWM_{Max} is the maximum PWM value.

The minimum force applied by the device was 0N, while the maximum force that the sleeve could generate was 5.1N.

All data recorded by the microprocessor during the experimental attempts was transferred to a PC via a USB serial cable and read and displayed in real-time via PuTTY application. Once each run was completed the application was closed and the data saved as a .txt file.

Ethical Considerations

The study was approved by the Informatics Research Ethics Board of the University of Edinburgh (2019/23785). Written consent was obtained from all participants prior to any experimentation.

Participant Selection

Participant inclusion criteria were being able-bodied and over the age of eighteen. Exclusion criteria were having a musculoskeletal disorder or prior experience with myoelectric control.

Experimental Setup

Figure 1 and **Supplementary Figure 1** depict the experimental setup. The electronic circuit, the Y-bridge and the biological hand were all positioned on the table. The myoelectric prosthetic hand was clamped on the edge of the table so that the experimental object it grasped was unsupported. The prosthesis used in this study was a six degree-of-freedom Nexus Hand (COVVI, Leeds, UK). However, only one degree of freedom was used as this allowed optimal replication of the grasping motion. Participants operated the prosthesis using two 50 Hz Össur surface electrodes with built-in EMG signal amplifiers and philtres (Össur, Reykjavik, Iceland). One electrode was adhered to the skin over the forearm extensor digitorum communis muscle group whist the other over the flexor digitorum superficial is muscle group. The experimental object was a 295 ml plastic tumbler cup. When the grasp force applied was >4.5N, the cup broke, producing a distinctive, loud noise. After each breakage, the cup was replaced with a new one.

Experimental Protocol

This study was a crossover randomised within-participant controlled trial. It began with a 5-min-long training phase consisting of guided familiarisation with myoelectric control. Next was the experimental phase made up of three experimental tasks that were performed under three feedback conditions (**Figure 2**). All participants completed all tasks under all conditions.

Sensory Feedback Conditions

Under the *visual feedback condition*, the participants did not wear the haptic sleeve and had their vision unobstructed. Thus, it was employed as a control condition. Under the *visual plus haptic feedback condition*, participants could still see but they also received mechanotactile feedback through the haptic sleeve. Under the *haptic feedback condition*, participants still received mechanotactile feedback but this time their vision was disabled by a blindfold. Any incidental auditory feedback was attenuated with the use of white noise-emitting headphones that participants wore at all times.

Experimental Tasks

Under each condition, participants had to perform fifteen 10-s-long repetitions of each task. In Task 1, *normal grasp*, the participants were instructed to grasp the experimental object with the myoelectric prosthesis so that the object neither breaks nor drops. Task 2, *strong grasp*, was the same as Task 1 but the instruction was to grasp the object as strongly as possible without breaking it. Task 3, *weak grasp*, was again the same as Task 1 but the command was to grasp the object as lightly as possible without dropping it. The purpose of the varying grasp strengths was to assess the impact of haptic feedback on grip force adjustment.

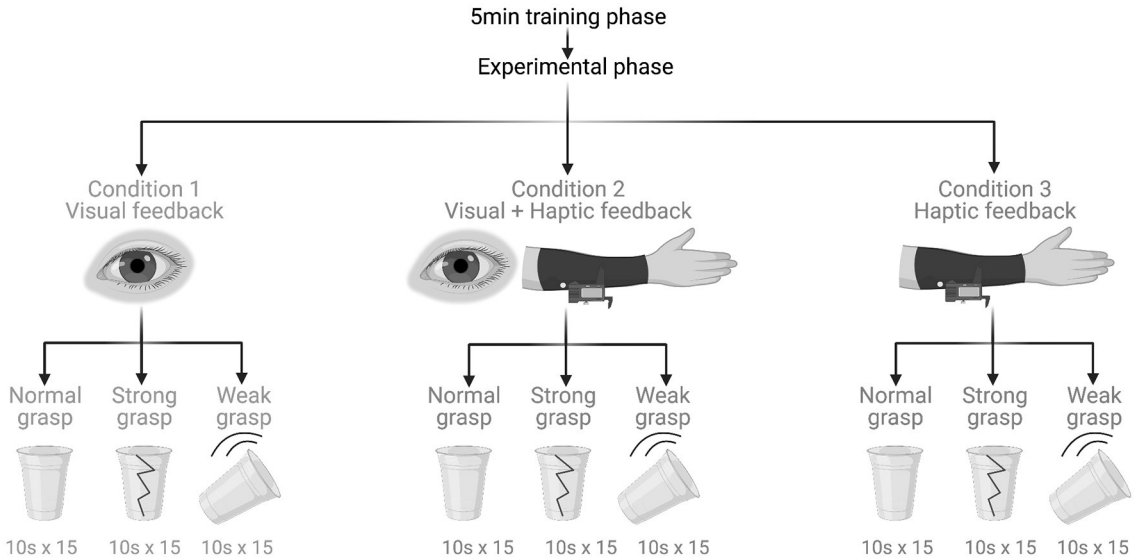

FIGURE 1 | Experimental protocol and tasks. Created with BioRender.com.

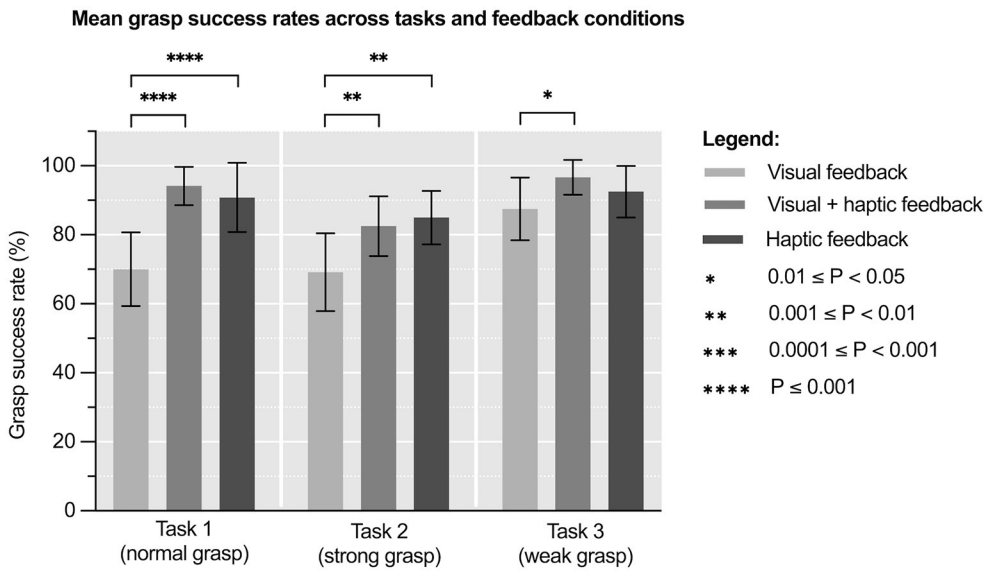

FIGURE 2 | Mean grasp success rates (%) across tasks and feedback conditions. Error bars represent standard deviation. $N = 8$.

Randomisation

Simple randomisation was performed to obtain a unique sequence of feedback conditions and tasks for each participant. The sequences were generated in Research Randomizer (Social Psychology Network, USA) and the participants were blinded to their allocated sequence. The aim was to reduce confounding of the results by learning effects.

Outcome Measures

The primary outcome measure was grasp success rate, expressed as a percentage of successful attempts. A successful attempt was defined as neither breaking nor dropping the experimental object. The secondary outcome measure was energy expenditure of grasping, equal to the indefinite integral of the EMG curve.

Sample Size Calculation

Sample size calculation was performed based on preliminary primary outcome results for the first four participants with a mean of 75% grasp success rate and standard deviation of 9.97 with visual feedback alone, compared to a mean 98.25% and standard deviation of 3.03 with haptic feedback alone. Glass' *delta* between these two groups demonstrated an effect size of 2.33.

Using this effect size, with a power of 80% and two-sided α level of 0.05, a desired sample size of eight participants was calculated using G*Power 3.1 (HHU, Düsseldorf, Germany).

Statistical Analysis

All statistical analyses were conducted in Prism 9.1.0 (GraphPad, California, USA). The threshold for statistical significance was adopted at $p < 0.05$. Shapiro-Wilk test of normality was performed first and showed all data were parametric. To determine if there was significance between groups, two-way analysis of variance (ANOVA) with repeated measures were conducted as there were two factors influencing the data (feedback condition and task). Partial eta squared (η_p^2) was calculated as an effect size measure of any significant results. *Post-hoc* Tukey's test was used to further characterise any statistical significance.

RESULTS

Study Demographics

In total, eight volunteers were recruited between March and April 2021. All of them met inclusion criteria and none were excluded. Hence, all volunteers were randomised; they completed all experimental tasks and were included in the analyses. **Supplementary Table 1** summarises their demographic characteristics. All experiments were conducted in April 2021 at the Scottish Microelectronics Centre, Edinburgh, UK.

Grasp Success Rate

Figure 3 and **Supplementary Table 2** show mean grasp success rates across feedback conditions and tasks. One way repeated measures ANOVA of these results revealed significant variation in grasp success rates under different feedback conditions in Tasks 1 & 2, but not during Task 3: Task 1 [$F_{(1.91,13.4)} = 48.5$, $p < 0.0001$], Task 2 [$F_{(1.25,8.75)} = 11.5$, $p = 0.006$] and Task 3 [$F_{(1.99,14.0)} = 3.47$, $p = 0.06$]. The effect size of this variation was $\eta_p^2 = 0.87$ (95% CI: 0.70–0.91) for Task 1, $\eta_p^2 = 0.62$ (95% CI: 0.24–0.73) for Task 2 and $\eta_p^2 = 0.42$ (95% CI: 0.04–0.59) for Task 3. This means that ∼87, 62, and 42% of variability in the results, respectively, can be attributed to feedback condition.

A *post hoc* Tukey's test showed that the mean grasp success rates under *visual plus haptic feedback condition* were significantly higher compared to those under *visual feedback condition* in all tasks: Task 1 (+34.6%, $p < 0.0001$), Task 2 (+19.2%, $p = 0.006$), and Task 3 (+10.5%, $p = 0.017$). It also showed that the mean grasp success rates under *haptic feedback condition* were significantly higher compared to those under *visual feedback condition* in Task 1 (+29.7%, $p < 0.0001$) and Task 2 (+22.8%, $p = 0.0015$). No significant differences were found between mean grasp success rates under *visual plus haptic* and *haptic feedback condition* in any of the tasks.

Energy Expenditure

Figure 4 represents mean EMG signal traces during all grasping attempts in respective tasks. These curves are timelines of participants' electromyographic activity (43). Mean areas under the EMG curves are presented in **Supplementary Table 3**. One way repeated measures ANOVA of these results revealed significant variation in mean areas under the EMG curves under different feedback conditions in all tasks: Task 1 [$F_{(1.31,9.18)} = 9.545$, $p < 0.009$], Task 2 [$F_{(1.82,12.8)} = 6.36$, $p < 0.01$), and Task 3 [$F_{(1.20,8.41)} = 9.51$, $p < 0.01$]. The effect size of this variation was $\eta_p^2 = 0.31$ (95% CI: 0.24–0.37) for Task 1, $\eta_p^2 = 0.12$ (95% CI: 0.07–0.17) for Task 2 and $\eta_p^2 = 0.24$ (95% CI: 0.17–0.30) for Task 3. This means that feedback condition accounts for ∼31, 12, and 24% of variability in the results across Task 1, 2, and 3, respectively.

A *post hoc* Tukey's test demonstrated that the *haptic feedback condition* was associated with significantly lower mean energy expenditure compared to the *visual feedback condition* in all tasks: Task 1 (−36.7%, $p < 0.0001$), Task 2 (−18.1%, $p < 0.0001$), and Task 3 (−22.4%, $p < 0.0001$). It also showed that the *visual plus haptic feedback condition* correlated with significantly lower mean energy expenditure compared to the *visual feedback condition* in Task 1 (−31.8%, $p < 0.0001$) and Task 3 (−8.7%, $p = 0.0003$).

DISCUSSION

Key Findings and Interpretation

This study was the first to adapt and test a wearable sleeve that integrates haptic feedback across the prosthetic socket as a method of mechanotactile feedback restoration. It showed that using the device correlates with higher grasp success rate and lower muscle energy expenditure, which is consistent with the initial hypothesis. These findings are clinically relevant. Difficulty grasping and muscle fatigue are some of the most important factors contributing to high prosthesis abandonment rate (12). Minimising their impact could increase prostheses function, potentially elevating amputees' overall quality of life.

Haptic Feedback Correlates With Higher Grasp Success Rate

Conditions including haptic feedback were found to be associated with significantly higher grasp success rates in all tasks (**Figure 3**). Additionally, feedback condition accounted for most variability in Task 1 and 2, suggesting a possible causative relationship. Correlation between non-invasive tactile feedback restoration and higher grasp success rates is well-established in bionic literature. Studies on electrotactile (16), vibrotactile (17), and mechanotactile (18) feedback all report the same trend. It is proposed that better grasp success rates result from participants utilising the feedback to better control the force they are applying (44). Another interesting finding was no significant difference in grasp success rates under *visual plus haptic* and *haptic feedback condition* in any of the tasks. There are two important implications to this. Firstly, it suggests that haptic feedback reduced participants' reliance on vision, a desirable phenomenon documented in other studies on tactile feedback (10, 11). Secondly, it might mean that the device's feedback delay is equal to, or even shorter than, visual feedback delay of 250 ms (45), making haptic feedback quick enough to be readily used (46).

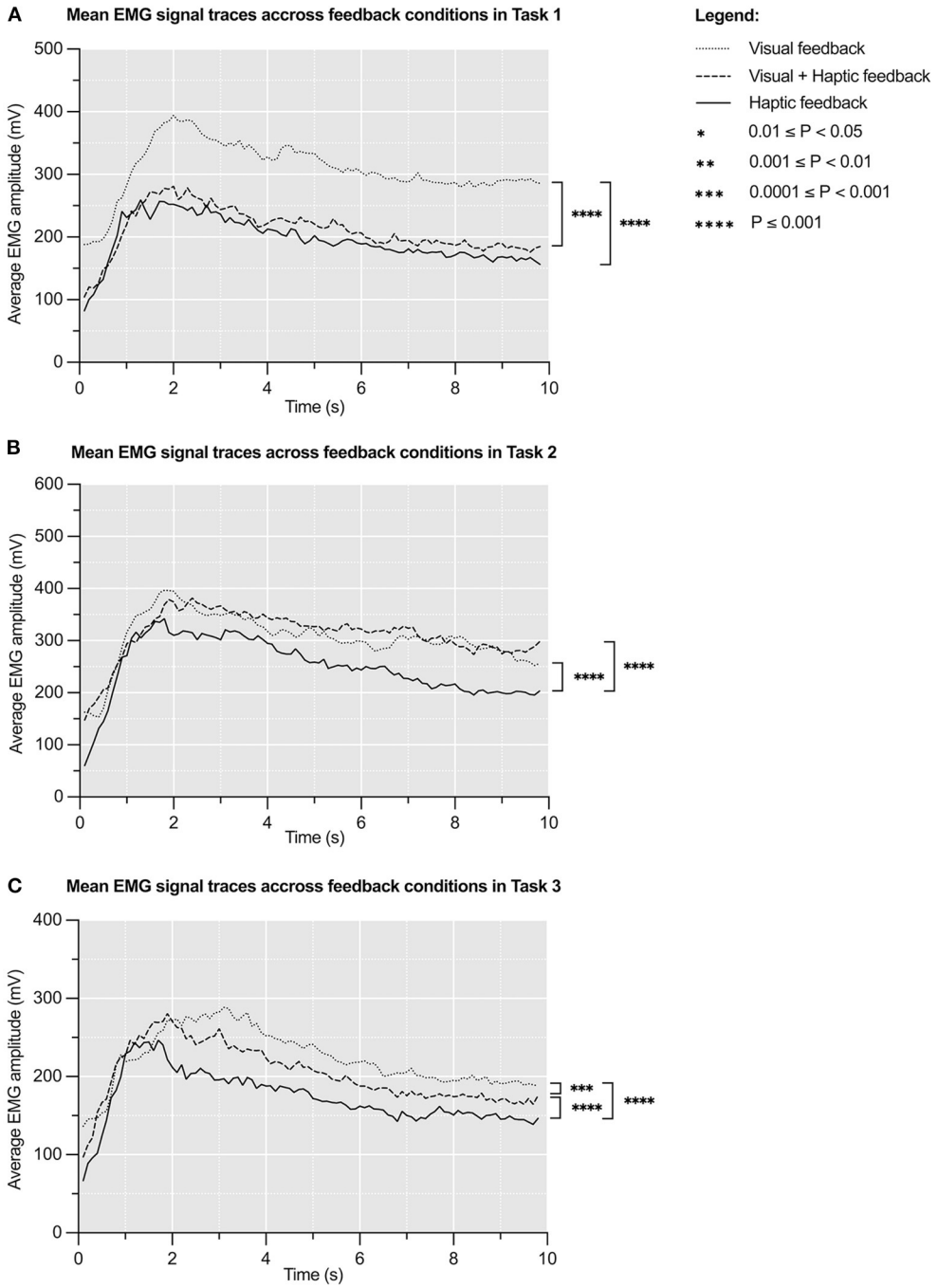

FIGURE 3 | Mean EMG signal (mV) traces over 10s across feedback conditions in Task 1 **(A)**, Task 2 **(B)**, and Task 3 **(C)**. The initial large peak corresponds to grasping the experimental object and the latter plateau corresponds to sustained grip. $N = 8$.

Haptic Feedback Correlates With Lower Energy Expenditure

Addition of haptic feedback was also noted to be correlated with significantly lower energy expenditure of grasping in all tasks, when the averaged EMG data was investigated across all trials (**Figure 4**). Moreover, it was estimated to be responsible for 12–31% of the variability in the results, suggesting a possible cause-and-effect relationship. It is challenging to compare this finding to existing literature as no previous study has calculated energy expenditure to assess sensory feedback restoration. Even studies that use EMG signals as a feedback modality (47, 48) do not report any outcomes in terms of energy expenditure. The reduction in energy expenditure demonstrated in this study can be explained by the closed loop motor control theory (1).

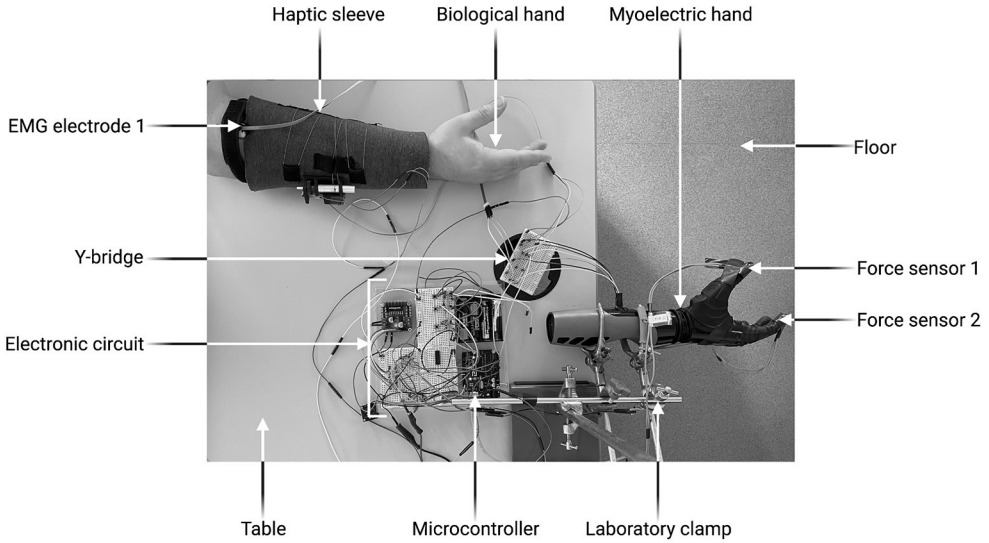

FIGURE 4 | Experimental set-up. Created with BioRender.com.

According to the theory, sensory feedback is constantly used to achieve economy of movement and the lowest possible energy expenditure (2). Additionally, in prosthetic control systems, sensory feedback can partially rectify the inherent uncertainty of the feedforward myoelectric control, improving the overall motor control (5). Thus, it is possible that mechanotactile feedback provided by the sleeve closed participants' motor control loops effectively enough to allow for motor control and energy expenditure to be optimised.

Strengths

Simultaneous EMG Signal Recording and Use

Muscle EMG traces are considered one of the most accurate methods of metabolic energy expenditure estimation for individual muscles and muscle groups (49). Measuring energy expenditure is of high importance in prosthetic research as it correlates with physical fatigue of prosthesis use (50). Fatigue, if excessive, leads to prosthesis abandonment (12, 51). Previous studies on tactile feedback restoration struggled to reliably record EMG signals due to inability to simultaneously record EMG signals and use them for myoelectric control (43). A Y-bridge, as used in this study, circumvents this issue by splitting the EMG signal registered by the electrodes into two independent channels. The technique is simple yet reliable and can be easily incorporated into future studies of this kind.

Experimental Sequence Randomisation

Haptic feedback use is characterised by a learning curve whereby functional outcomes improve with practise (52, 53). To reduce the confounding effects of learning on this study's results, simple randomisation of experimental task sequence was performed. Not only did this distribute the learning bias across feedback conditions and tasks, but also hindered learning by eliminating predictability (54). Despite each participant performing a unique sequence of tasks, significant changes were found in favour of haptic feedback, which corroborates the study's internal validity.

Adequate Sample Size and Power

The required sample size was met and therefore the study is 80% powered for the primary outcome to keep the probability of Type II error at 0.2 and, thus achieves a statistically sound balance with Type I error probability of 0.05 (55). However, as this study particularly looked at the impact of the tasks on EMG activity as a marker of energy expenditure, and there was limited previous data to calculate the sample size, our calculations were based on the pilot data of 4 participants. Consequently, while the significant results obtained in this investigation may reflect a true effect, further work with amputee participants should help corroborate these findings.

Limitations

Lack of Amputee Participants

Due to the concurrent COVID-19 pandemic, recruiting amputee participants was made impossible. Considering the situation, the study was adapted to accommodate able-bodied participants. However, this was a suboptimal solution due to a number of significant differences between a residual limb and a healthy hand. During the amputation procedure, certain muscles and nerves are partially or completely removed, impeding subsequent EMG signal generation and making it more variable compared to able-bodied counterparts (56). Several changes may occur after the amputation, too. These include residual limb muscle atrophy, phantom limb pain or sensations, as well as contracture and neuromata formation (57). As a result, amputee participants may find it more difficult to not only generate EMG signals sufficient for myoelectric control, but also perceive and interpret the mechanotactile feedback provided by the haptic sleeve. Thus, the study's generalisability to amputee population is compromised.

Incomplete Natural Sensory Feedback Disablement

One common criticism of using able-bodied participants in studies assessing sensory feedback restoration is that these volunteers have their natural sensory feedback intact, which can confound the results (28). Although this study experimental setup meant that natural tactile feedback was negligible (a healthy hand did not touch the experimental object), participants still had their proprioception intact (they could move their biological hands). As a consequence, they were able to understand their hands' position even with their vision disabled, which could arguably result in better performance compared to amputees. To fully disable all sensory feedback, peripheral nerve blocks (58) or inflatable cuffs (59) have been used in similar studies in the field. However, the benefit of using these methods has to be balanced against their invasiveness and painfulness, as well as the requirement for additional ethical considerations.

Implications for Future Research and Clinical Practise

Future research should aim to further develop and test the haptic sleeve. Firstly, the device should be adapted for amputee use and assessed in the target user group. Additionally, the effects of the haptic sleeve need to be studied on a greater range of manipulative tasks. For that, established clinical tests assessing user performance in myoelectric control can be used, such as the Southampton Hand Assessment Procedure or the Action Research Arm Test (60). Any significant effects can later be studied over time to characterise the learning curve.

In order to successfully introduce the haptic sleeve onto the market and into clinical practise, it needs to be made portable. The current prototypic electronic circuit can be miniaturised into a compact control platform which will easily fit within the sleeve. If future studies corroborate clinical benefits of using the haptic sleeve, the device may become an integral element of the rehabilitation process after an upper limb amputation. Thanks to the sleeve, future amputees might be able to achieve better prosthetic function, which may translate into greater prosthesis embodiment, reduced phantom limb pain, enhanced quality of life and wider job opportunities.

CONCLUSION

This study demonstrates that a haptic sleeve can be an effective tool for mechanotactile feedback restoration. Use of the device correlates with significantly higher grasp success rates and significantly lower energy expenditure of grasping in healthy volunteers. These findings are likely due to the haptic sleeve improving control of the applied force, decreasing reliance on vision and closing the motor control loop. Further research into the area is warranted and should focus on adapting the device for amputee use and improving its portability. With these enhancements, the haptic sleeve may help amputees recover more function and improve their quality of life.

DATA AVAILABILITY STATEMENT

The raw data supporting the conclusions of this article will be made available by the authors, without undue reservation.

AUTHOR CONTRIBUTIONS

AR conceived the study, with VB, AM, SV, and AS contributing to the design and implementation of the work. VB and AM collected the data and analysed the data with AR. All authors contributed to drafting and reviewing the manuscript before final approval for submission.

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SUPPLEMENTARY MATERIAL

The Supplementary Material for this article can be found online at: https://www.frontiersin.org/articles/10.3389/fresc.2022.806479/full#supplementary-material

Supplementary Figure 1 | Data flow through the experimental set-up. Created with BioRender.com.

Supplementary Table 1 | Demographic characteristics of the study population. Numbers in brackets represent standard deviation. BMI, Body Mass Index. $N = 8$.

Supplementary Table 2 | Mean grasp success rates (%) across tasks and feedback conditions. Numbers in brackets represent standard deviation. $N = 8$.

Supplementary Table 3 | Mean energy expenditures (mV) across tasks and feedback conditions. Numbers in brackets represent standard deviation. $N = 8$.

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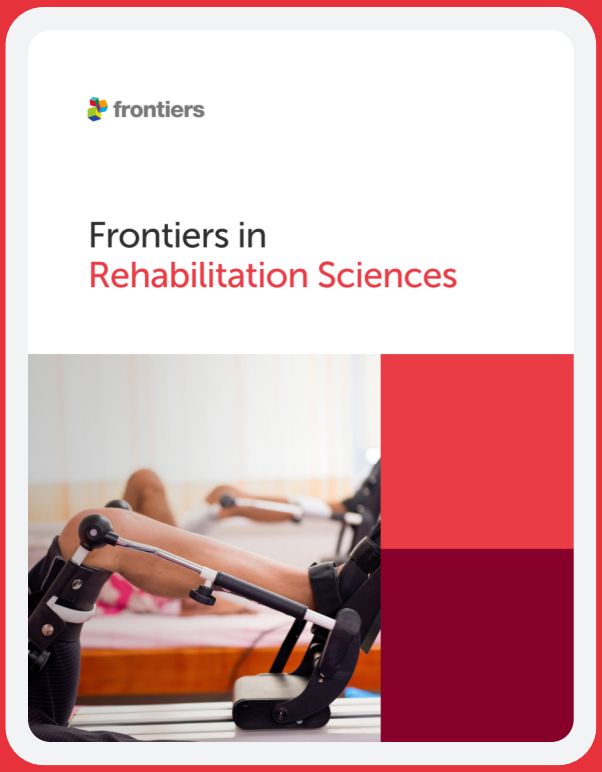

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