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Transverse rotation and longitudinal translation during prosthetic gait—A literature review

Martin Twiste, MSc; Shyam Rithalia, PhD
Directorate of Prosthetics and Orthotics, University of Salford, Manchester, United Kingdom; School of Health Care Professions, University of Salford, United Kingdom

Abstract—Improved technology allows for more accurate gait analysis to increase awareness of nonoptimized prosthetic gait patterns and for the manufacture of sophisticated prosthetic components to improve nonoptimized gait patterns. However, prescriptions are often based on intuition rather than rigorous research findings for evidence-based practice. The number of studies found in the literature that are based on prosthetic research regarding transverse rotation and longitudinal translation is small when compared to topics regarding other types of movements. Some design criteria for prosthetic components described in those studies that permit transverse rotation and longitudinal translation can be found in current designs. However, little research has been conducted to establish their effectiveness on the gait parameters and residual limb. This literature review is an investigation into these motions between the socket and the prosthetic foot, with particular reference to gait characteristics and prosthetic design criteria.

Key words: artificial limbs, displacement, gait, leg prosthesis, locomotion, movement, pressure, rotation.

INTRODUCTION

One of the simplest lower-limb prosthetic designs was the peg leg, which consisted of a residual limb interface and a post to provide a connection to the ground [1]. Adding a foot improved both aesthetic and functional aspects by allowing the amputee to wear a shoe with the prosthesis. It concealed the prosthesis and enlarged the contact area with the ground, thus increasing stability because of a change in action of ground reaction forces. The resultant design therefore consisted of a socket and a prosthetic foot, with a post serving as a connection between the socket and the prosthetic foot rather than the socket and the ground. Although materials and suspension systems have changed since then, current designs are still based on this type of composition and some have additional features, including a conversion of the rigid post into an articulated connector. For reasons of simplicity and durability, most of these articulations provided motion predominantly in the sagittal plane and were particularly beneficial for high levels of amputation, because of the loss of proximal joints and the subsequent need for longer prostheses.

Abbreviations: BSK = Blatchford Stabilised Knee, SACH = solid ankle cushioned heel, SFESK = Stanceflex Endolite Stabilised Knee, TT = telescopic and torsional, VSP = vertical shock pylon.

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Address all correspondence and requests for reprints to Mr. Martin Twiste, Directorate of Prosthetics and Orthotics, Brian Blatchford Building, University of Salford, Manchester M5 4WT, UK; email: m.twiste@salford.ac.uk.

Clearly, the need for articulated prostheses was apparent, and joints were also introduced in prostheses for low levels of amputation. Ongoing research led to the realization that motions at anatomical joints may occur in more than just one plane. Because of new materials and improved technology, prosthetic components could be manufactured that permitted motion other than purely in the sagittal plane to simulate motion at anatomical joints. Despite changes in lower-limb prosthetic designs, one feature that had to be incorporated and could not be disposed of is the residual limb-socket interface. However, this is a source for skin problems because of excessive angular and linear motion of the residual limb within the socket, causing high pressures and shear forces. These problems stress even more so the need for articulations within prostheses to accommodate motion of the residual limb and to minimize associated problems. Findings from the literature with respect to articulations that allow movement between the socket and the prosthetic foot in the transverse plane and combined coronal and sagittal plane will be discussed in this paper.

DEFINITION OF NOMENCLATURE

Terminologies encountered in the literature for motion in the transverse plane and combined coronal and sagittal plane, as well as for prosthetic components that permit these motions, vary and some are even ambiguous or misleading. Therefore, appropriate terminology will be required to provide a clear description of these types of motions and prosthetic components.

Combined coronal and sagittal plane motion may be regarded as longitudinal motion, which also identifies transverse plane motion because this occurs in a plane perpendicular to the longitudinal axis. To address the problem associated with excessive angular and linear motion of the residual limb within the socket as mentioned earlier, depending on their function, one therefore can regard articulations that may accommodate such motion as transverse rotation adapters, longitudinal translation adapters, or combined transverse rotation and longitudinal translation adapters.

CONSIDERATIONS FOR PROSTHETIC MOTION

One of the major problems associated with the design of prostheses is to create an interface, which allows ground reaction forces that usually act through the anatomical foot in able-bodied persons, to be transmitted via the prosthesis onto the residual limb. Because of possible hypersensitivity of the residual limb caused by the amputation and a surface area that is distally somewhat smaller than that of the anatomical foot, ground reaction forces have to be distributed over as large an area of the residual limb as possible. However, only a small amount of forces acts perpendicularly on the skin, causing the majority of forces to create shear stresses. With possible orientations in circumferential and longitudinal directions, these forces can be excessive in magnitude because of shock impact and may exacerbate shear stresses. Angular and linear motion of the residual limb can therefore be regarded as the cause for circumferentially and longitudinally directed shear stresses, which implies that if motion of the socket in these directions were permitted by introducing transverse rotation and longitudinal translation, the resulting stresses may be reduced.

Another difficulty encountered in the design of artificial limbs is the assimilation of prosthetic gait to that of nonpathological, able-bodied persons to restore both aesthetic and functional aspects of the missing limb [2]. From the aesthetic point of view, prostheses that appear, in terms of exterior modeling and gait performance, like an anatomical limb can be considered an accurate replica of the absent limb. With respect to functional maximization, a number of authors have reported on differences in prosthetic designs and their influence on the gait parameters, whereby minimization of energy consumption seemed to be considered one of the most advantageous effects on prosthetic gait [3,4]. It is debatable, though, whether a gait pattern similar to that of nonpathological, able-bodied persons is the most appropriate type for amputees. At this stage, however, this will be assumed to be the case. Consequently, considering that, as mentioned above, transverse rotation and longitudinal translation are assumed to be beneficial for the reduction of shear stresses acting on the residual limb, it may be advantageous to provide this motion to amputees by implementing appropriate prosthetic components.

MOTION IN ANATOMICAL LOWER LIMB

Transverse Rotation Between Anatomical Segments

A study by Levens et al. revealed that transverse rotation of the tibia, femur, and pelvis occurs throughout swing and stance phase [5]. However, no analyses were given for possible implications of that motion on the gait except that it can be regarded as "an important factor in the ease and rhythm of walking." Lafortune et al. found that during peak knee flexion, the amount of internal rotation of the tibia was greatest when shoes with a valgus wedge were used and smallest when shoes with a varus wedge were used [6]. However, the presence or the type of wedge neither affected the transverse rotation of the tibia during the remaining part of stance phase nor the tibiofemoral joint throughout stance phase. From their findings, Lafortune et al. concluded that changes in the magnitude of transverse rotation of the tibia must therefore affect transverse rotation at the hip [6]. However, no evidence was given to substantiate their assumption and changes in transverse rotation of the tibia could possibly trigger compensatory motion further proximal to the hip. This hypothesis was supported by Nester et al. who found that, despite significant effects on the rearfoot complex motion, medially and laterally wedged foot orthoses have a negligible influence on transverse rotation at the knee, hip, and pelvis [7]. Because of movements of skin relative to the underlying bone [8], Steinman traction pins more accurately identify anatomical landmarks than skin markers do and were used in the study by Lamoureux and Radcliffe and Levens et al. [9,5].

However, drastic changes in the angular relationship of the lower-limb segments with respect to the cameras can cause errors because of perspective and parallax. Levens et al. made corrections to compensate for these errors [5], but because of differences in technology available when these two studies were conducted, the accuracy of the study by Lafortune et al. is likely to be greater [6]. Nevertheless, both studies demonstrated that transverse rotation of the tibia is characterized by two main events, which involve progressive internal rotation during the initial part of stance phase, followed by progressive external rotation during the remaining part of stance phase. Unlike Levens et al. whose results involved transverse rotation of the tibia, femur, and pelvis and at both intersegmental joints throughout the gait cycle, Lamoureux and Radcliffe limited their results to transverse rotation of the tibia and at the tibiofemoral joint for the stance phase only [5,9]. Therefore, despite limited information in the study by Lafortune et al. [6], the data by Levens et al. [5] should only be used as a general guide for gross motion rather than for a detailed motion analysis.

During swing phase, without a mobile, anatomical structure that allows relative motion between the shank and the foot, transverse rotation of the shank would cause transverse rotation of the foot. But because of the presence of ground reaction forces during stance phase, lack of transverse rotation of the foot as a result of friction with the ground would also limit transverse rotation of the shank if intersegmental mobility would not exist. Manter studied such structures and established the orientations of axes and the types of motion occurring at the subtalar and transverse tarsal joint [10]. Building on these concepts, Subotnick and McPoil and Knecht elaborated, amongst others, on the intricate action of pronation and supination and the relationship between these triplanar motions and transverse rotation of the shank [11,12]. In turn, Reischl et al. established that magnitude and temporal characteristics of peak transverse rotation of the tibia and femur could not be predicted from the magnitude and temporal characteristics of peak foot pronation [13]. However, motions of body segments were referred to the global reference system, which Nester described as ambiguous because motions between segments are, in a clinical situation, generally referred to as joint motion, thus to the anatomical reference system [14]. In this study, Nester demonstrated that the magnitude and temporal characteristics of transverse plane motion at the hip and knee could not be directly correlated with rearfoot complex motion. Nevertheless, the authors of both studies agreed that, because of the interaction between joints, the extent of variability between subjects might contribute to a lack of correlation between intersegmental motion but that a relationship must obviously still exist that prevents the anatomical foot from rotating together with proximal segments.

Lamoureux and Radcliffe investigated the effects of a locked and unlocked transverse rotation adapter on transfemoral gait, thus simulating the absence and presence of such a device, respectively [9]. They not only found that the amount of transverse rotation of the pelvis was greater with the transverse rotation adapter unlocked compared to when it was locked, but they also found that the relative amount of transverse rotation between pelvis and socket increased and the torque at the adapter decreased. Internal and external rotation of the socket with respect to the pelvis reached a maximum at the beginning of swing phase and during mid stance phase, respectively. However, the study by Levens et al. [5], which is used as a general guide only, as suggested previously, shows that internal and external rotation of the femur with respect to the pelvis reached a maximum at the beginning of stance phase and at the beginning of swing phase, respectively. Therefore, with the residual limb embedded in the socket, the incorporation of a transverse rotation adapter does not cause the socket to undergo transverse rotation similar in amount or pattern to that of the femur during nonpathological gait of able-bodied persons. The reason for this could be that the amount or pattern of transverse rotation of the femur during nonpathological gait of able-bodied persons is different to that of the residual femur in amputees. Alternatively, Lamoureux and Radcliffe suggested that contractions of the rectus femoris muscle and gluteal muscles may apply a transverse rotation moment on the socket because of bulging of those muscles during hip flexion and extension [9]. Whether the characteristics of soft tissue properties are the sole influence on transverse rotation of the socket or whether the encapsulation of the bones inside the socket also affects motion of the socket remains open. What is shown, though, is that nonpathological gait of able-bodied persons is clearly characterized by transverse rotation and that this motion still has an effect on the prosthesis during amputee gait, which demonstrates the need for a transverse rotation adapter. However, considering that, during swing phase, because of the absence of ground reaction forces and therefore the absence of friction between the prosthetic foot and the ground, the socket is not restricted from undergoing transverse rotation, which is why a transverse rotation adapter can be considered necessary for stance phase only.

Longitudinal Translation Between Anatomical Segments

Extending the duration of collision time is a means of dissipating energy and helps reduce shock impact [15], a process based on a number of mechanisms in the anatomical lower limb, including soft tissues, bones, and motion at the joints [16]. This process changes the force-time ratio by allowing the body's center of mass to decelerate gradually rather than abruptly because of a relative shortening of the lower limbs.

Angular displacement between segments of the lower limbs can be regarded as one of the causes for such shortenings and can be observed at both the ankle and knee. Lafortune et al. demonstrated that following the heel strike, the knee joint flexes before returning to nearly full extension during the second half of stance phase [6]. Linear translation is another cause for a relative shortening of the lower limbs. Responsible for a large reduction of the acceleration experienced by the lower limb [16], the heel pad represents one of the most important shock absorbers because of its spongy characteristics. Because its absorbency depends on its thickness [17], wear because of repetitive impact may compromise the absorbing characteristics by triggering degenerative changes. However, additional shock absorbers including shock absorbing shoes and shoe inserts can be worn, which help to prevent or compensate for excessive wear of anatomical shock absorbers [15,18,19]. Also, the foot can be considered a shock absorber as a whole because of the relative displacement of the bones as a result of the elastic properties of tendons, ligaments, and muscles, which allow the foot to deform and by doing so to store and release energy [20]. Regardless of whether angular displacement or longitudinal translation takes place between segments of the anatomical lower limb, the fact that the body possesses numerous mechanisms that allow such motions demonstrates that shock impact needs to be reduced in prostheses also.

MOTION IN PROSTHETIC LOWER LIMB

Transverse Rotation Between Prosthetic Segments

Multiaxial mobility is a design characteristic incorporated in some prosthetic feet to simulate motion at the joints of the anatomical foot. Some designs are based on compression and deflection of elastic materials and others on mechanisms with rotation axles. However, despite no obvious technical restrictions and despite the possibility to permit motion in any direction, the use of transverse rotation in prosthetic feet is limited.

According to Thomsen [21], an early design of a prosthetic foot permitting transverse rotation was patented in 1921 and was later incorporated in the design of the Roesser-Gummiblock-Gelenkfuß.* More recent developments include the Multiflex foot and Greissinger foot, which allow motion in all three planes.† Boonstra et al. investigated motion of the Multiflex foot [22], but measured angular displacement in the sagittal and coronal

*Roesser, Essen, Germany.

†Multiflex foot: Chas. A. Blatchford and Sons Ltd, Basingstoke, UK; Greissinger foot: Otto Bock Orthopädische Industrie GmbH und Co., Duderstadt, Germany.

plane only. Buchgold investigated the design characteristics of and maintenance procedure for the Multiflex foot but, despite describing transverse rotation as an unusual feature in prosthetic feet, the author did not elaborate any further on this matter [23]. Edelstein described some features of the Multiflex foot and Greissinger foot and explained that motion in all three planes not only helps those feet to conform to the ground but also reduces shear stresses transmitted to the residual limb [24]. Esquenazi and Torres listed some design characteristics of a number of prosthetic feet, which permit transverse rotation, but failed to elaborate on possible benefits of such motion on the gait [25]. Incorporating a mechanism for transverse rotation in the prosthetic foot rather than providing an additional mechanism to provide such motion proximal to the foot may be advantageous because of a possible reduction in space requirements and weight. However, this may be achieved at the expense of reduced durability and a foot with such a mechanism incorporated may not be the most suitable type for a particular amputee. The choice of suitable prosthetic components therefore is greater when deciding on separate parts.

Mulby and Radcliffe suggested that the socket during transfemoral gait undergoes 3° of internal and 10° of external rotation with up to 75 lb in (approximately 8.48 N•m) of torque acting on the lower limb [26]. Amputees with skin problems or with a range of motion exceeding 5° of transverse rotation between the pelvis and the foot were defined as those who would benefit from such a design, but no explanations were given as to how any of the figures were derived. Staros and Peizer described a transverse rotation adapter, which permitted 7° of rotation and, unlike current designs, in one direction only [27].* From this finding, one can assume that the authors were not referring strictly to unidirectional motion but to either internal or external rotation, with the device subsequently returning to its neutral state. Whether transverse rotation in only one direction provided any advantage over devices with bidirectional motion was not clarified.

Housing two ball bearings and an elastomer torsion spring, the "UC-BL shank axial rotation device" was tested by Lamoureux and Radcliffe on transfemoral amputees who, with the device incorporated in the prosthesis, experienced a reduction in skin problems [9].† Despite positive results, the authors suggested that transtibial amputees might be less in need of such a device because of a normal hip and the resultant freedom for transverse rotation. First, however, no evidence was given that motion at the hip in transtibial amputees is what they described as "normal." Second, the authors considered transverse rotation adapters less critical for transtibial amputees because they probably assumed that the resultant freedom for transverse rotation because of a "normal hip" counteracts and therefore neutralizes to a certain extent the rotation occurring at the pelvis. It can be seen from the study by Levens et al. [5], which is used as a general guide only, as previously discussed, that transverse rotation of the femur was shown to be greater than and in phase with transverse rotation of the pelvis. This demonstrates that transverse rotation of the femur does not counteract transverse rotation of the pelvis but that it occurs in addition to it. Therefore, if the assumption is true that transverse rotation at the hip is similar in transtibial amputees to that of non-pathological gait of able-bodied persons, this stresses the need for a transverse rotation adapter for transtibial amputees despite this being considered less critical by Lamoureux and Radcliffe [9]. Schmidl recorded the absolute values for transverse rotation during transtibial and transfemoral gait [28], but without relating these data to temporal gait characteristics. It was shown, however, that the sum of internal and external rotation was approximately 14° to 15° and similar for both levels of amputation and greater in symmetry with respect to the neutral position in transtibial amputees. Therefore, despite Lamoureux and Radcliffe's assumptions that transtibial amputees may be less in need of a transverse rotation adapter [9], for reasons previously explained, findings by Schmidl demonstrate that the need for such a device is of similar importance for transtibial and transfemoral amputees [28].

To design transverse rotation adapters with suitable characteristics, one must consider the type of mechanical features that need to be incorporated. Lamoureux and Radcliffe suggested a small and lightweight mechanism that would fit into a variety of prostheses and to reduce the inertia of the limb [9]. A dampening mechanism was proposed so as to prevent excessive vibrations because of the transverse rotation adapter returning to its neutral state too quickly. Transverse rotation of up to 20° in each direction was considered a sufficient range with mechanical stops and a unit to return the adapter to its neutral state that can be customized for each amputee, whereby no reasons were given for these considerations. Also, despite their

*Lord Corp. of Erie, Pennsylvania, USA.

†UC-BL shank axial rotation device: Biomechanics Laboratory, University of California, Berkeley, USA.

recommendations with regards to adjustments of the return unit, a resistance of about 0.23 N•m/° was suggested. This was, without substantiation, considered low enough so as not to exceed the friction at the residual limb-socket interface and high enough to permit the adapter to return to its neutral state. The authors also proposed asymmetrical, nonlinear forces from the return unit and rotation that does not occur purely in the transverse plane. Kaphingst analyzed a variety of resistances to transverse rotation [29], including 0.15 N•m/°, 0.23 N•m/° and 3.9 N•m/°, and considered the first and the last figure as quite "soft" and quite "hard," respectively. However, neither were the sources of those figures defined nor were the assessments of those figures substantiated. This author also suggested a design, which allows the transverse rotation adapter to be converted into a rigid adapter, if required, by manually locking it in its neutral state for security on different grounds.

As to the location of a transverse rotation adapter within transfemoral and hip disarticulation prostheses, Knoche considered it to be irrelevant whether transverse rotation occurs proximally or distally [30], but without backing up this assumption scientifically. More recent studies revealed that in transfemoral prostheses [31,32], the amounts of transverse rotation between the socket and pelvis were larger with the transverse rotation adapter incorporated than without it, as it was the case in the study by Lamoureux and Radcliffe [9]. However, with a transverse rotation adapter incorporated in the shin section and later in the thigh section, the tests were inconclusive regarding the location of a transverse rotation adapter, possibly because of lack of statistical power having studied the gait of two subjects only.

Prescribing transverse rotation adapters can be confusing and misleading when the decision is made based on commercial literature often designed according to a marketing strategy rather than validated scientific evidence. Some authors reported on specialized activities to indicate the prescription of transverse rotation adapters and emphasized the importance of such devices for torsional movements, including golf [33,34]. This is also an activity probably used as one of the most frequently encountered prescription guidelines in commercial literature but, in most cases, lacking scientific backup. Quesada et al. and VanNess et al. analyzed the performance of golf swings and found (amongst other deviations from motion patterns in nonpathological, able-bodied persons) a reduction in the amount of hip rotation and club swing and speed of hip and shoulder rotation [35,36].

Longitudinal Translation Between Prosthetic Segments

Prosthetic knees may be flexed during swing phase but remain fully extended throughout stance phase for sufficient stability from the prosthesis [37–39]. This, however, not only prevents shock absorption because of the lack of compliance from the prosthesis but also prevents the prosthetic gait pattern from being assimilated to nonpathological gait of able-bodied persons. A weight-activated mechanism the "bouncy knee" stabilizes the joint while a rubber bush in which its single axle is embedded permits some degree of rotation and therefore controlled knee flexion [40,41]. This was based on a modified Blatchford Stabilised Knee (BSK) and later led to a design with similar features, the current Stanceflex Endolite Stabilised Knee (SFESK).* Polycentric mechanisms represent alternative solutions, e.g., 3R60 with part of the linkage arrangement designed to permit knee flexion, during which the instantaneous center of rotation is shifted further posteriorly so as to maintain stability of the knee [42].† With the 3R60 knee, shortening of the shin section during flexion, as is characteristic for polycentric knees with a certain geometry [43], occurs in addition to a relative shortening of the whole prosthesis as with the bouncy knee. Therefore, with both knees undergoing the same amount of flexion, the overall relative shortening and shock absorption should theoretically be less with the bouncy knee than with the 3R60 knee. Also, prosthetic feet that permit motion in the sagittal plane by allowing it to plantarflex or to simulate plantarflexion by compressing the heel dampen impact at heel strike because of the shortening of the length of the prosthesis as with the anatomical ankle and heel pad [24,25]. However, it is debatable whether the amount of dampening is sufficient without some degree of knee flexion as in joints described previously.

The Terry Fox jogging prosthesis (a spring-loaded telescopic unit incorporated in the shin section) facilitates longitudinal translation and is designed not only to absorb shock impact during compression of the spring but also to

*BSK and SFESK: Chas. A. Blatchford and Sons Ltd, Basingstoke, UK.

†3R60: Otto Bock Orthopädische Industrie GmbH und Co., Duderstadt, Germany.

release the energy stored later during stance phase [2,44]. However, DiAngelo found that because of a spring mechanism with an inappropriate amount of resistance and a pneumatic damper [2], which seemed to counteract the decompression of the spring, the desired extent of shock absorption and energy release was not achieved. Flex-Foot incorporated the concept of energy storage and shock absorption in the design of prosthetic feet.* In addition to carbon-fiber leaf spring technology as used in all of their Flex-Foot designs, the Reflex VSP (vertical shock pylon) foot comprises a telescopic shin section with an external carbon-fiber leaf spring to control longitudinal translation. Neither permitting nor restricting longitudinal translation in this foot greatly affected the gait parameters. This could be due to the lack of statistical power with this study having recruited only two subjects [45,46]. In turn, with the use of a SACH (solid ankle cushioned heel) foot, a Flex-Foot, and a Reflex VSP foot during transtibial gait, the results favored the latter with regards to improving gait efficiency and reducing energy expenditure and exercise intensity [47–49]. However, no explanations were given as to what type of SACH or Flex-Foot was being used. Yack et al. demonstrated that with transtibial amputees ascending stairs using the same prosthetic feet, the work done by the hip on the amputated side was greatest using the SACH foot [50], which indicates that energy storage can have a positive effect on prosthetic gait.

Fergason and Boone also elaborated on the importance of longitudinal translation to reduce shock impact by using a telescopic shin section [33]. Although these authors pointed out that the amount of longitudinal translation is adjustable in the majority of such devices, no examples were given. Instead, what is adjustable in most adapters examined for this literature review is the resistance to longitudinal translation, which, in turn, will consequently affect the amount of travel, providing that the same amount of force is applied. Changing the amount of travel without altering the resistance to longitudinal translation can have detrimental effects. For instance, a reduction in the amount of travel possible may cause longitudinal translation to halt abruptly when the mechanical stops inside the adapter are hit, making the use of such devices pointless because shock impact would still occur.

*Flex-Foot, Inc., Aliso Viejo, USA.

Transverse Rotation and Longitudinal Translation Between Prosthetic Segments

Staros and Peizer reported on a device (S.P.T. Limb) that provided, in addition to transverse rotation, longitudinal translation to simulate knee flexion [27].† The authors gave no descriptions on the type of design or the actual effects of such a device on the gait. In turn, research into the telescopic and torsional (TT) pylon was conducted by a number of authors, but its effects on gait were interpreted primarily because of longitudinal translation rather than transverse rotation.‡ Buckley et al. reported on the gait of six transtibial amputees who, with the TT pylon incorporated in their prostheses, experienced a reduction in energy expenditure when traveling at 130 percent and 160 percent of their normal walking velocity [3]. Despite the lack of significant findings encountered during normal walking velocity, transverse rotation was not being considered as the source for the effects found at higher speeds. Whatever were the actual reasons for those findings, three of the subjects who benefited most from the TT pylon also had this device incorporated in their own prosthesis, which is why those authors considered it possible that increased familiarity with the device may maximize its effects on the gait.

Gard and Konz found that incorporating a TT pylon in a transtibial prosthesis caused a decrease in the initial, vertical ground reaction forces [51]. Although this effect may be the result of longitudinal translation, the authors did not elaborate on whether transverse rotation of the device was possible or whether it was locked, which is why assuming that only longitudinal translation was responsible for such effects would be speculating. Ross and McLaren analyzed the gait of 10 transfemoral amputees with two different lower-limb setups using a TT pylon with longitudinal translation permitted and with this motion restricted [52]. However, the results were unclear, in that the authors reported a change in the vector profiles but did not elaborate on what measurements these profiles represent. Also, as in the study by Gard and Konz [51], Ross and McLaren did not clarify whether transverse rotation was possible or not, which is why an interpretation of the results is similarly speculative [52]. Using the OS1 and US1 adapter, the study by Stauf was based on gait analysis regarding both transverse rotation

†Een-Holmgren, Uppsala, Sweden.

‡Chas. A. Blatchford & Sons Ltd, Basingstoke, UK.

and longitudinal translation, unlike previously described studies [53].[*] Tests were conducted with only one transfemoral and one transtibial amputee, which demonstrates lack of statistical power. Also, in addition to insufficient clarity concerning the description of technical details for the collection and analysis of the data, the relationship between the torque around a vertical axis and transverse rotation appears to be based on a static rather than dynamic situation, which may falsify the results. In turn, with regards to longitudinal translation, the results showed a reduction of the force-time ratio for the early parts of stance phase by 25 and 20 percent for the transfemoral and transtibial amputee, respectively, which indicate positive shock-absorbing characteristics of those adapters.

Current Designs

Currently, available adapters that permit transverse rotation, longitudinal translation, or both are typically based on a proximal and distal housing, with a return unit located in between. The return unit will be distorted or compressed while the two housings are being rotated or telescopically displaced in opposite directions to one another during transverse rotation or during longitudinal translation, respectively. Because of the elastic properties of the return unit, distorting or compressing it generates resistance to those motions and allows such adapters to return to their neutral or fully elongated state upon removal of opposing forces. The mechanical representation of such units is based, amongst others, on elastomers, coil springs, or as previously mentioned, on leaf springs, whereby the density, thickness, and width determine the resistance to that motion.[†]

Future Developments

To maximize the accuracy of future gait analysis, it would be beneficial to test transverse rotation and longitudinal translation separately, together, and one and not the other. Also, in addition to using force plates and computerized motion analysis systems, measuring the residual limb-socket interface forces may provide further insights into the effects of prosthetic devices that allow such motion to occur.

[*]medipro Technik, Bayreuth, Germany, designed by Century XXII Innovations, Inc., Jackson, USA.

[†]Century XXII Innovations, Inc., Jackson, USA.

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