

## Chapter 46

# Transfemoral Amputation: Prosthetic Management

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### Abstract

*Clinicians have long strived to create optimal transfemoral prosthetic designs that will not only enhance the user's ability to ambulate but also will be a functional element of the individual's life. Although there have been many advancements in materials, socket designs, and components, there has been little research to help quantify how the individuals that use these prosthetics devices can best be served. It is helpful to explore clinical considerations and anticipated outcomes when creating transfemoral prosthetic devices. Prosthesis use is affected by many factors, including energy expenditure, body image, voluntary control within a transfemoral prosthetic system, socket fit and design, component selection, and alignment.*

**Keywords:** prosthetics; transfemoral alignment; transfemoral prosthetic socket; transfemoral prosthetic suspension

### Introduction

Amputations at the transfemoral level account for approximately 19% of the approximately 1.6 million individuals in the United States who are currently living with an amputation.<sup>1,3</sup> Statistics from 2004 reported that 31% of all major amputations were performed at the transfemoral level,<sup>4,5</sup> with new evidence showing a decrease in the number of transfemoral amputations performed each year.<sup>2</sup> There also is evidence that individuals who have undergone amputation are living longer and will require prosthetics services throughout their lives.<sup>1,6</sup> In 2014, certified prosthetic practitioners spent more than 25% of their time caring for patients with a transfemoral amputation.<sup>7</sup>

Those in the field of prosthetics have a long history of involvement with transfemoral prosthetic socket design and construction, with the first patents awarded in England in 1790 and the first US patent for a transfemoral artificial limb given in 1846.<sup>8-12</sup> However, prosthetists still do not have universal

clinical standards of practice for device creation, fit, suspension, and alignment. Throughout history, transfemoral design and fabrication techniques have been passed down from mentors to protégés, with no formal instructional courses available in the United States until 1949 when the University of California at Berkeley introduced a short course in transfemoral design of a suction socket. In the 1950s, several universities began formal education programs in prosthetics, with each school creating their own laboratory manuals and design iterations.<sup>13</sup> The prevalent design at that time was the German transfemoral quadrilateral socket, which used skin suction suspension.<sup>14,15</sup> In the 1980s, the first ischial containment manual emerged and was quickly adapted by other institutions, although each institution implemented design iterations. As of 2014, there were 11 accredited institutions offering master level education for prosthetic and orthotic practitioners in the United States, with each institution offering differing theories and practical

implementation techniques for transfemoral socket design, suspension, and clinical application.

One rationale for the differing designs may be the variability observed in the anatomy, size, and length of transfemoral residual limbs, as well as the level of voluntary control the individual possesses. It is accepted that no single design is appropriate for every individual with a transfemoral amputation. Accordingly, variations in the clinical applications of formalized training have led numerous practitioners to create unique styles and techniques.<sup>9,10,16,17</sup> These variations provide practitioners with the ability to adapt a transfemoral socket to best meet the needs and goals of an individual patient.

The common clinical goals and considerations that guide rehabilitation professionals through this patient-specific process are discussed in this chapter along with an overview of current transfemoral socket designs and the implications of suspension, alignment, and biomechanical considerations in evaluating, fabricating, and fitting transfemoral prostheses.

### Clinical Considerations and Anticipated Outcomes

Although prosthetic devices will never truly replace a missing limb, certain clinical considerations must be addressed, irrespective of which socket or suspension design is chosen. The transfemoral prosthetic system must balance function, comfort, and appearance both dynamically and statically.<sup>8,11,12</sup> To create the most appropriate plan, the treating team must consider energy expenditure, body image, the user's level

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of voluntary control, and the fit of the prosthetic socket. In implementing the treatment plan, the team must determine socket construction and design, the degree and complexity of the suspension system, the appropriate components, alignment considerations, and outcome measures.

### Energy Expenditure

Energy expenditure for a transfemoral amputee is of great concern. The effort required to ambulate with a prosthetic device at this level is dependent on the weight of the device, the quality of fit, the degree of suspension, the accuracy of alignment, and functional characteristics of the chosen components.<sup>17-21</sup> If any one of these factors is not properly addressed, the individual using a transfemoral prosthesis will exhibit higher levels of energy expenditure during ambulation than are necessary. Increased energy expenditure is accompanied by an increase in the rate of oxygen consumption and an associated elevation in heart rate. An elevated heart rate can, in turn, lower the user's self-selected walking speed and reduce gait efficiency.<sup>22</sup> For elderly individuals with a transfemoral prosthesis, the physical burden of ambulating with a prosthetic device may exceed their abilities, leading to a lower rate of prosthetic use.<sup>20,23</sup> Knowing that ambulation with transfemoral prosthetic devices requires high levels of energy, practitioners must create treatment plans that meet the individual's needs and goals with an acceptable burden level.

### Body Image

Body image and appearance when using a transfemoral prosthesis are complex considerations and should be addressed within the treatment plan. It is important to realize that appearance and self-image can be a cosmetic as well as a functional concern. An acceptable appearance and the ability of the user to

integrate with peers plays a large role in an individual's positive adaptation to his or her altered body image and psychosocial adjustment.<sup>24</sup> Body image anxiety increases depression, reduces perceived quality of life, lowers self-esteem, reduces participation in physical activity, and lowers overall satisfaction.<sup>25,26</sup> The prosthetist must create a device to maximize the confidence of the user through optimal fit, suspension, function, and alignment symmetry, as well as an acceptable energy expenditure.<sup>27</sup> There is a growing trend toward user participation in aesthetic choices, including realistic silicone covers, water transfers, or three-dimensionally printed cosmeses. These choices may help the user feel more involved with the creation of his or her prosthesis and thus increase device acceptance.<sup>24,28</sup>

### Effect of Voluntary Control

Functional ambulatory goals will be defined by the individual's ability or potential to control the transfemoral prosthetic device. This is commonly known as the level of voluntary control.<sup>9,10,14,29,30</sup> Because the user will not have direct musculoskeletal control of the prosthetic knee and foot, a determination of his or her potential voluntary control is an important consideration in determining the socket style, interface, suspension, and components used. Factors that determine the degree of voluntary control include residual limb length, positional awareness in space, active range of motion, muscle strength, and the ultimate ability to manipulate the limb in a controlled and deliberate manner. When voluntary control is limited, the rehabilitation team should design prosthetic systems that focus on prosthetic support and patient safety rather than function and performance. In contrast, enhanced voluntary control allows for the design of a more dynamic prosthesis. The degree of voluntary control also plays an integral role in

component choice and alignment considerations.

### Fit of the Prosthetic Socket

The ideal goal for any prosthetic device is for the user to feel that the device is part of his or her body. Irrespective of the socket design, an optimal fit should be intimate to the contours of the residual limb and assist the user in controlling the prosthesis. Beyond these basic criteria, an optimal fit of a transfemoral prosthetic socket is poorly defined and has not been standardized. However, if users do not feel that they have control of the socket, they likely will not fully use the prosthetic device.<sup>28,31</sup>

Radcliffe<sup>14</sup> suggested that the primary goals of a transfemoral prosthesis are to achieve comfort in weight bearing, provide a narrow base of support in standing and walking, and accomplish the swing phase of gait in a manner that is as close to normal as possible. The fit and orientation of the socket are paramount in achieving these goals. The socket must be donned in the correct orientation with respect to the user's line of progression, must match the volume of the residual limb, and must create an environment of total contact without causing impingement or discomfort. The socket also should provide adequate stability in the sagittal, coronal, and transverse planes throughout the gait cycle.

### Importance of Orientation

When donning the socket, the orientation of the socket must match the user's residual limb and adjacent bony structures. If the socket is malaligned, the device will rotate and cause undue pressure on the limb or the pelvis. To properly integrate the limb within the socket, the individual should be instructed regarding socket orientation as it relates to his or her anatomy. This anatomic reference differs for the varying socket designs but must be addressed, especially in the initial and subsequent fittings of the device.

**Importance of Total Contact Socket Fit**

There are various techniques to assess whether the volume of the socket matches the volume of the residual limb. Most techniques rely on a combination of visual verification through a clear diagnostic interface and a determination of internal socket pressure through visual examination, tactile probes, or electronic pressure sensors. Irrespective of the technique, it is imperative that pressures are balanced and can be tolerated by the user.<sup>16,32</sup> The prosthetist should ensure that all areas within the socket make contact because lack of contact may result in edema, socket migration, and compromised control of the prosthesis.

The tissues proximal to the trim lines must be free from impingement throughout gait and while seated. Tissue bulging over the proximal trim lines can lead to skin breakdown, edema, subdermal cysts, blisters, irritation, and discomfort.<sup>33</sup> Similarly, there must be adequate relief for the bony structures within the socket. Pressure on the ischial tuberosity, ascending pubic ramus, adductor longus tendon, greater trochanter, or distal femur can lead to socket rotation, pain, gait deviations, or rejection of the prosthetic device.<sup>34</sup>

**Socket Stability**

Stability of the transfemoral prosthetic socket on the limb is vital to the control of the device. The prosthetist will make a clinical determination on the type of socket design based on the individual's level of voluntary control and the stability required. Individuals with greater levels of voluntary control are less dependent on socket modification, component choice, and alignment accommodations to control unwanted socket displacement during ambulation. Excessive motion of the transfemoral prosthetic socket on the residual limb in the sagittal, coronal, and transverse planes can lead to increased energy

expenditure, gait deviations, and dissatisfaction with the prosthesis.<sup>23,35,36</sup>

**Sagittal Plane**

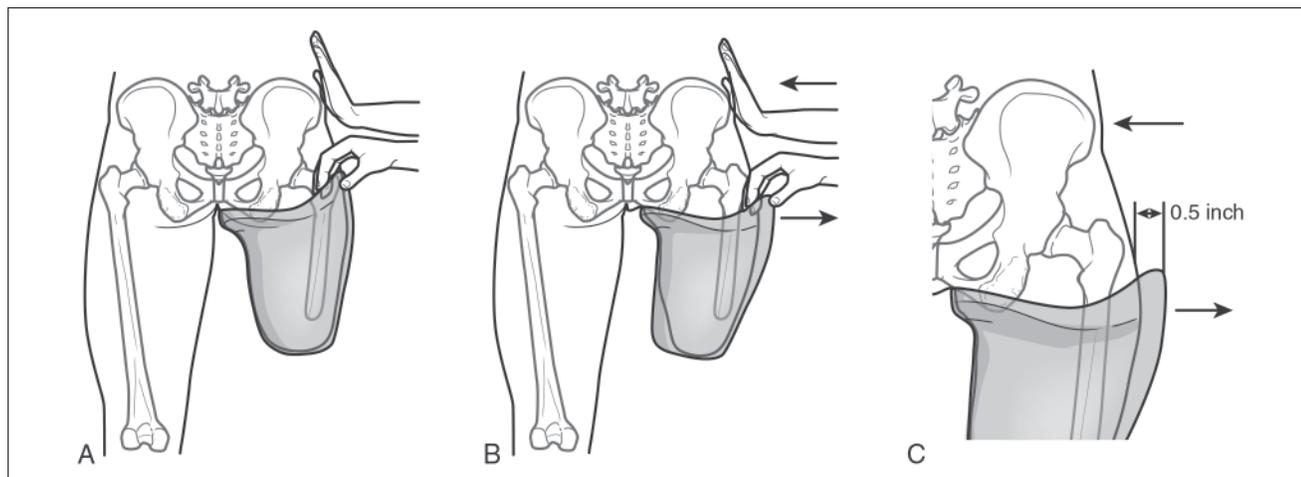
The principles of prosthetic control in the sagittal plane are best considered in the early stance phase of the gait cycle. As the prosthetic foot contacts the floor, the ground reaction force quickly moves posterior to the mechanical knee joint center and creates an external knee flexion moment that will cause the prosthetic knee to buckle if it is not adequately controlled by the user. The ipsilateral hip extensor musculature of the individual must fire, pulling the residual femur and the prosthetic socket posteriorly to create a counterextension moment and stabilize the mechanical knee.<sup>37</sup> Importantly, the residual femur must be adequately stabilized within the socket before the actions of the hip extensors can be translated through the prosthesis to act on the ground. In the absence of such femoral stabilization, the contractions of the hip extensor muscles are less effective, and the ability to control the prosthetic knee is compromised, causing the individual to compensate with a reduction in step length, a slower cadence, or an anterior shift in body weight. All of these compensatory actions increase energy expenditure.

Prosthetic control in the sagittal plane should also be considered in late stance. During this phase of the gait cycle, the individual must engage the hip flexors to drive the prosthetic socket anteriorly. This hip flexion action creates prosthetic knee flexion, thereby lifting the overall prosthesis off the ground to initiate the swing phase. Inadequate femoral stabilization may delay the execution of this action, resulting in a loss of control of the prosthetic knee and potential compromise of its function. The individual will likely display a shortened step length, reduced speed of ambulation, and a lack of confidence with the prosthetic device.<sup>28,31</sup>

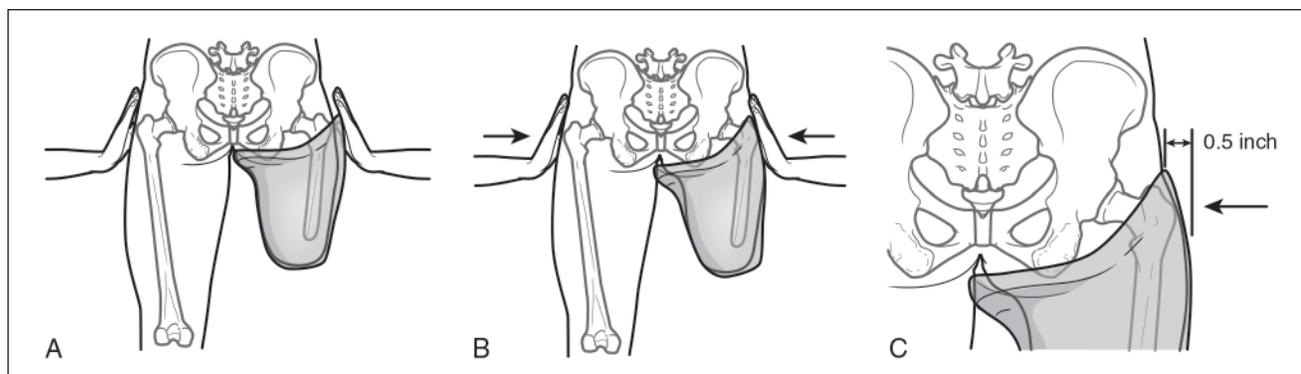
**Coronal Plane**

In the coronal plane, prosthetic control is critical in limiting the movement of the torso laterally over the prosthetic device during the single-limb support phase of the gait cycle. This compensatory lateral movement over the prosthesis is one of the most common prosthetic gait deviations seen in the user of a transfemoral prosthesis.<sup>38</sup> Unless a hip abduction contracture is present, the residual femur should be placed in an adducted position equal to the contralateral femur. This position ensures the efficient firing of the hip abductor muscles on the amputated side, which limits contralateral pelvic drop and associated lateral trunk bending. This is accomplished by fitting a flattened lateral socket wall that is countered by a sufficiently high medial socket wall aligned in the correct angle of femoral adduction.<sup>14,30,39,40</sup>

During the initial fitting of a transfemoral socket, the proximal coronal instability of the socket can be easily determined by performing the lateral and the medial displacement tests. For both of these assessments, the prosthesis user must be standing safely within parallel bars. To perform the lateral displacement test, the prosthetist places one hand on the proximal lateral brim of the transfemoral socket while the other hand is placed on the prosthesis user's ipsilateral iliac crest. Gently but firmly, the prosthetist then pushes medially on the iliac crest while also pulling laterally on the proximal brim of the socket. If the socket displaces more than 0.5 inch (1.27 cm) from the residual limb during this static test, the socket may also displace laterally during single-limb stance in gait. This lateral displacement often suggests coronal instability in the socket, and it can cause the individual to experience excessive proximal medial pressures on his or her residual limb. A compensatory lateral shift of the torso may be adopted to restore coronal stability and reduce these pressures (**Figure 1**).



**Figure 1** Illustrations of the steps in the lateral displacement test. After donning, the socket is aligned with the line of progression and checked to ensure a total contact fit and a level pelvis. The prosthetist then can test for lateral displacement of the socket on the limb. **A**, One hand is used to grasp the proximal edge of the socket while the other hand is placed on the ipsilateral iliac crest to provide a counterforce and stabilization. **B**, The proximal socket is pulled laterally until displacement stops. **C**, The ideal amount of displacement is 0.5 inch (1.27 cm) measured from the skin to the socket wall. If the displacement is greater than 0.5 inch (1.27 cm), the transfemoral socket likely will be unstable in the coronal plane during single-limb support.



**Figure 2** Illustrations of the steps in the medial displacement test. After donning, the socket is aligned with the line of progression and checked to ensure a total contact fit and a level pelvis. The prosthetist then can test for medial displacement of the socket on the limb. **A**, One hand is placed over the proximal-lateral aspect of the socket and the other hand is placed over the greater trochanter on the contralateral side. **B**, Both hands are used for medial compression until socket displacement ceases. **C**, The ideal amount of displacement is 0.5 inch (1.27 cm) from the starting point. If the displacement is greater than 0.5 inch (1.27 cm), the transfemoral socket likely will be unstable for the user in the coronal plane during single-limb support.

The medial displacement test is performed with medial, simultaneous compression of the proximolateral aspect of the socket and the greater trochanter of the contralateral limb. Medial socket displacement of more than 0.5 inch (1.27 cm) may suggest that either the mediolateral dimension of the transfemoral socket or its overall volume is too large. Alternatively, the prosthesis user may not possess enough voluntary control to resist the lateral forces created during single-limb support<sup>40-43</sup> (**Figure 2**).

#### *Transverse Plane*

Transverse stability, observed in the swing and early stance phases of gait, also is dependent on both the level of the individual's voluntary control and the optimal fit of the transfemoral socket. During the evaluation of the residual limb, the strength of its subcutaneous tissue and musculature should be assessed to help determine if the individual can control the normal transverse plane motions of gait, including internal rotational motions during the swing phase and external

rotations during early stance. If either the muscle or the underlying connective tissues are found to be inadequate, the individual will not be able to voluntarily control these forces, and the socket may rotate on the limb. In such cases, either targeted socket modifications or external components are needed to aid in controlling transverse rotation.

If the individual has adequate voluntary control but still demonstrates whip-type gait deviations or excessive socket rotation, these problems may be caused by a suboptimal socket fit, with

**Table 1** Transfemoral Socket Construction

Primary Socket Construction				
Construction	Example	Primary Indication	Major Advantages	Chief Limitations
Hard socket (Integrated rigid inner socket and outer frame)		Mature limbs Firm limbs Situations that allow for reduced trim line height Minimal wall thickness	Simple design Durability Easy to clean Solid construction that will not alter over time	Comfort with rigid proximal trim lines Hard surface when sitting Adhered tissue, invaginations or sensitive bony areas may not be accommodated
Removable flexible inner socket; rigid outer frame (Flexible inner socket can be removed from outer frame)		Any limb shape Flexibility in design Dynamic muscle movement Relief areas for sensitive tissue Proximal soft-tissue support	Adjustable Allows for dynamic muscle movement and relief for sensitive tissue while maintaining a total contact fit Comfort in sitting Fenestrations can be created while retaining socket strength	Inner socket may change shape over time Durability
Emerging Socket Construction				
Removable, flexible inner socket; rigid outer frame; dynamic panels (Panels are adjustable to change compression on inner flexible socket)		Volume fluctuations Progressive pressure in specific areas User adjustability	Dynamically alters inner socket shape and compression felt on limb in specific areas Compression from panels is user adjustable	Lengthy fabrication process More maintenance
Flexible socket; embedded rigid frame (Rigid frame is laminated between layers of flexible material)		Mature limbs High levels of voluntary control Dynamic muscle movement	Flexible interface Soft proximal brim Comfort in sitting Soft outer surface Rigid embedded frame supports socket with minimal surface area	Lengthy fabrication process Fragile Limited adjustability Heavier than other construction types

volumetric incongruities exerting the largest influence on rotational control. To reduce transverse rotation, socket fit must be optimized to match the individual's limb volume or accommodations must be made for muscle contractions.

### Primary Socket Designs

#### Socket Construction

The hard socket and the flexible inner socket with a rigid frame are the two

general classifications of socket construction for transfemoral prostheses. The flexible inner socket has two variations that are gaining in popularity: the flexible inner socket with dynamic panels and the flexible socket with an embedded rigid frame (**Table 1**).

#### Hard Sockets

Although all sockets are constructed around a positive model of the

prosthesis user's limb, a hard socket is a single-walled, static socket that is designed to be in direct contact with either the user's skin or an interface such as a roll-on gel liner or prosthetic sock. The advantages of a hard socket include its simplicity, thin-walled construction, durability, and ability to be easily cleaned and maintained. Because this socket option offers little padding and cannot absorb the shear forces generated

between the limb and the socket walls, it is intended for limbs with stable volume, firm tissue, and fair to good skin sensation. This socket construction is generally contraindicated in individuals with adhered scar tissue, invaginations, and sensitive bony prominences; these prosthesis users require a more forgiving design.<sup>33</sup>

A transfemoral hard socket is typically fabricated from a carbon fiber or a rigid thermoplastic material with no fenestrations or cutouts. This socket construction does not readily allow for fluctuations in residual limb size, so an optimal initial fit is crucial. Similarly, the prosthesis user with a hard socket must be diligent in volume management to maintain a total contact fit.

#### ***Flexible Inner Sockets***

The second type of socket construction incorporates a flexible inner socket capable of elastic movements and a rigid outer frame for stability. This design can be in direct contact with user's skin or an interface such as a roll-on gel liner or a prosthetic sock. The main advantage of this design is that the inner flexible socket, which is made from a silicone-based thermoplastic material, allows for both volumetric and localized fitting accommodations. In addition, when the proximal trim lines of the rigid frame are lowered, the flexibility of the proximal inner socket dramatically increases the user's comfort. The brim of the inner socket contains the proximal tissue, but it allows for elastic movement around sensitive bony areas such as the ischial tuberosity, the ascending pubic ramus, and the anterior superior iliac spine while sitting. The flexible inner socket also allows for increased proprioceptive feedback when the rigid outer frame is cut away in strategic areas. When the rigid frame is fenestrated, the compliant material of the inner flexible socket is exposed to allow an individual to feel the surface he or she is sitting on or to have room for

residual musculature to expand during ambulation. Even with the frame cut away, the inner flexible socket contains the soft tissue and maintains the benefits of hydrostatic weight bearing. In the absence of the inner flexible socket, such fenestrations would allow the individual's skin to protrude from the openings, compromising both hydrostatic loading and pressure distribution during weight bearing and leading to localized window edema and skin breakdown.

One variation of a flexible inner socket with a rigid outer frame uses dynamic panels that can be adjusted to help regulate pressures within the socket. In this variation, instead of using open cutouts, the outer rigid frame is fabricated with free-floating panels that are connected by tensioning cords. In the fabrication process, hollow tubes are laminated into the frame in strategic locations. After the panels are cut out of the frame, the hollow tubes are exposed, allowing tensioning cords to be fed through the tubes in both the panels and the rigid frame. The cords can be tightened as needed to move the panels closer to the inner flexible socket, thereby increasing the overall compression felt by the user. The main advantage of this system is the ability of the user to change the shape and volume of the socket. When necessary, the dynamic panels can be loosened to allow bulbous limbs to enter the socket, and then compressed to create a total contact fit. The panels also can be adjusted on an activity-specific basis, such as relieving tension while sitting or kneeling or increasing compression for strenuous activities such as running.

An alternative design is a flexible socket with an embedded rigid frame. This design is the result of advances in materials and fabrication techniques. Similar to prosthetic systems popular before World War II in which a metal frame was housed within a flexible leather socket,<sup>8</sup> this modern variation uses rigid frames laminated within an otherwise flexible socket. This socket

construction option has been successfully used in upper limb prosthetic applications and is now beginning to be used with transfemoral prostheses.<sup>44</sup> The major advantages of this socket construction include its overall flexibility, reported comfort, minimal trim lines, and perceived improved control of the prosthesis.<sup>45</sup> Although this socket is heavier, more difficult to fabricate, and less durable than other types of sockets, the final product offers a very flexible system with confined, rigid support.

#### **Socket Designs**

The two primary socket designs for transfemoral prostheses are the ischial ramal containment and the subischial designs. Two subcategories of the ischial ramal containment design are the ischial containment (IC) and the ramal containment (RC) designs. Variations of the subischial design include the quadrilateral design and those designs that may incorporate the use of subatmospheric-assisted vacuum suspension (**Table 2**).

#### ***Ischial Containment Designs***

In current practice, the IC socket is the most commonly used design, with numerous iterations in both teaching and clinical practice. All variations of the IC socket have the common goal of providing mediolateral stability in single-limb support. This goal is achieved by using an intimately fitted socket with a narrow mediolateral dimension while encasing the medial aspect of the ischial tuberosity and ramus within the socket.<sup>30</sup> Most IC design variations use hydrostatic weight bearing rather than direct ischial weight bearing. In contrast to the quadrilateral socket, the medial IC wall is angled to match the ischial ramus angle of the prosthesis user rather than the line of progression. The IC socket can help minimize the lateral thrust of the socket during single-limb support by buttressing against bony aspects of the pelvis. If the orientation of the IC wall

**Table 2** Transfemoral Socket Designs

Design	Example	Primary Indication	Major Advantages	Chief Limitations
Ischial/ramal containment (The ischial-ramal complex is contained within the socket to provide stability in the coronal plane during single-limb support)	Ischial containment 	When enhanced coronal stability is requested User presents with lower levels of voluntary control Shorter residual limbs	Enhanced coronal stability with support from bony structure Proximal tissue contained inside of socket Amount of containment can be adjusted Potential for ischial weight bearing	Clinical experience required to create optimal fit High proximal trim line May inhibit hip range of motion
	Ramal containment 	Enhanced coronal stability with minimal trim lines	Enhanced coronal stability with support from bony structure Proximal tissue outside of socket Reduced proximal trim lines Enhanced hip range of motion with maintenance of coronal stability	Clinical experience required to create optimal fit Discomfort unless optimal fit around ascending ischial ramus is achieved Lengthy fitting process
Subischial (The ischium is not contained within the socket. Coronal stability during single-limb support is obtained by soft-tissue compression)	Quadrilateral 	Longer residual limbs User presents with higher levels of voluntary control Previous user	Ischial weight bearing	Clinical experience required to create optimal fit Minimal coronal stability Narrow anteroposterior dimension Predetermined rectangular socket shape
	Subischial, vacuum-assisted suspension 	Longer residual limbs High voluntary control High gadget tolerance Uses hydrostatic weight bearing	Reduced trim lines Excellent suspension that may enhance coronal stability issues due to intimate fit Amount of assisted vacuum suspension force can be regulated Hip range of motion not limited	Clinical experience required to create optimal fit Tissue proximal to the brim may be stressed Multistage donning process May be a bulky design

angle does not match the anatomic angle of the ischium and ascending ischial ramus, the prosthesis user will likely feel undo pressure and discomfort with this socket design.

The amount of IC is variable, with most designs initially containing the

ischium from 1 to 1.75 inches (2.54 to 4.445 cm) proximal to its distal aspect.<sup>29,30,41</sup> This amount of containment gives adequate mediolateral control while allowing for tissue around the ischium and ascending ischial ramus to aid in padding the sensitive bone.<sup>46</sup>

Sockets that incorporate more proximal degrees of IC typically have more proximal gluteal containment as well.

Some IC designs, such as the Marlo Anatomical Socket (Ortiz International), suggest that coronal stabilization can be achieved by limiting bony containment

to the ascending ischial ramus and lowering the other proximal trim lines of the socket.<sup>47</sup> This socket design allows greater range of motion about the hip and decreases some metabolic costs.<sup>18,20,47</sup> However, this type of design is difficult to fit and may be rejected because of localized pressures on the medial ramus. Achieving an optimal fit requires the prosthetist to have an elevated level of clinical skills.

Given the proximal intrusions of IC designs into perineum and bony elements of the pelvis, the common challenge of IC designs is determining the amount of actual bony support that can be tolerated by the individual user. IC designs tend to work well for individuals with shorter residual limbs or those who lack voluntary control of their adductor muscles. For individuals with longer residual limbs and high degrees of voluntary control, aggressive IC may be unnecessary, and a subsichial design might be more appropriate.<sup>19,43,48</sup>

### **Subsichial Designs**

The quadrilateral socket is a subsichial design that uses ischial weight bearing on its posterior brim with some degree of additional hydrostatic loading of the residual limb to support the individual's weight.<sup>14</sup> The ischium rests on the posteromedial aspect of the socket brim where it is held in place through anteroposterior socket compression. In contrast to IC socket designs, this anteroposterior tightness requires an increased mediolateral dimension to allow proximal soft tissue to enter the socket. However, this increased mediolateral dimension can create a lack of coronal support, leading to pressure on the perineum and common gait deviations such as lateral trunk flexion and a wide base of support. Also common with the quadrilateral design is difficulty in achieving adequate lateral support for the femoral shaft, which often leads to the reduced effectiveness of the gluteus medius to stabilize the pelvis in single-limb support. This lack of

femoral support was reported by Long<sup>39</sup> in the 1980s and led to the initiation of early IC designs.<sup>10,46</sup>

As the quadrilateral socket has declined in popularity, there has been growing interest in a subsichial design that uses hydrostatic weight bearing as does the IC socket but does not incorporate the ischium into the socket. This design was introduced in the 1960s by Redhead.<sup>16</sup> Although his work contributed to a better understanding of hydrostatic weight bearing, the design did not achieve widespread acceptance.

Current subsichial socket designs are based on the original concepts described by Redhead,<sup>16</sup> but they also incorporate a roll-on gel liner interface and assisted vacuum suspension. These subsichial sockets may be preferred over IC designs because of their lowered proximal trim lines.<sup>43,45</sup> In addition, there are suggested advantages of increased limb health, volume stabilization, reduced perspiration, and increased comfort.<sup>49</sup> Studies are needed to objectively prove these suggested advantages for varying limb lengths and levels of voluntary control, but the design appears to be a viable choice at this time.

### **Suspension Systems**

Total contact socket fit and adequate suspension throughout the entire gait cycle is necessary to ensure the confident use of a transfemoral prosthesis. During the stance phase of gait, total contact is maintained by the user's weight. During the swing phase, the inertia and weight of the prosthesis will displace the socket from the residual limb if the suspension system is inadequate. On taking the next step, the user will force the limb back into the socket, creating a piston-type motion. This displacement or pistoning of the limb within the socket, even if it is a few millimeters, can lead to loss of prosthetic control, skin irritation, overall socket discomfort, distal residual limb edema, and gait deviations.<sup>11,33</sup>

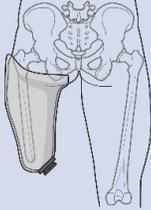
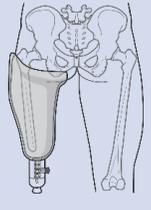
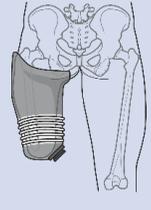
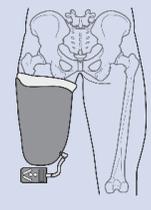
Because of the large amount of remaining compliant soft tissue, general residual limb shape, and minimal bony femoral anatomy, suspension is often quite difficult to achieve for an individual using a transfemoral prosthesis. Various forms of suspension have been attempted to minimize transfemoral socket displacement. Although all of these variations have merit, there is no evidence that supports a clinical standard for a single suspension system.<sup>33</sup> Therefore, it is imperative for the prosthetist to have a working knowledge of the various systems available and take into consideration the unique goals and characteristics of each transfemoral prosthesis user.

Current suspension systems being used at the transfemoral level are generally classified as subatmospheric, negative-pressure, and belt-type systems. Subatmospheric systems use some level of negative atmospheric pressure combined with surface tension to maintain the transfemoral socket on the residual limb. Subcategories of subatmospheric designs include skin-fit suction, roll-on liners with various locking mechanisms, roll-on liners with a hypobaric sealing membrane, and vacuum-assisted suspension. Belt-type systems use positive, superiorly directed forces created by a strapping system secured around the pelvis. Subcategories of belt systems include the Silesian belt, elastic belt suspension, and hip joint and pelvic belt suspension (**Table 3**).

### **Subatmospheric Suspension**

Subatmospheric suspension provided by skin suction or a roll-on gel liner is the most prevalent type of suspension design.<sup>33</sup> These systems work by combining friction with a negative pressure differential within the socket to maintain suspension of the prosthesis on the residual limb. In the literature, the term "suction" is used synonymously with the term "vacuum" when discussing

**Table 3** Transfemoral Suspension

Suspension Type	Example	Primary Indication	Major Advantage	Chief Limitations
Subatmospheric, negative pressure (Primary means of suspension used with most transfemoral prosthetic systems)	Skin-fit suction 	Whenever clinically feasible Mature limb	Increased proprioception between limb and prosthesis	Susceptible to volume changes Difficult to don Limited absorption of shear and impact forces felt on limb
	Roll-on gel liner with pin lock, lanyard, or magnet locking mechanism 	Firm limb Expect volume changes	Easy to don Secure suspension Allows for volume changes Absorption of shear and impact forces	Distal distraction can lead to edema Hand strength and dexterity required to apply liner Alignment and build height considerations
	Roll-on gel liner with hypobaric seal 	Firm limb Expect volume changes	Easy to don Can allow for volume changes Minimal distal distraction Absorption of shear and impact forces	Hand strength and dexterity required to apply liner
	Roll-on gel liner, vacuum-assisted suspension 	Mature limb Active Minimal socket displacement	Excellent suspension increases proprioception and control Minimal socket displacement decreases daily trauma to limb	Bulky Difficult to don Gel liner is fragile Gadget tolerance needed Maintenance

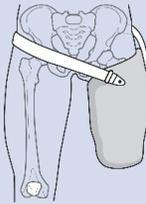
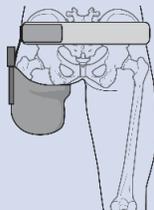
transfemoral suspension systems. A suction system has been defined as a subclass of subatmospheric socket systems that allows air to be expelled from a sealed socket while preventing air from entering the socket. However, the

internal pressure of the socket environment is not actively regulated.<sup>50</sup>

The suction necessary for these suspension systems can be created between the skin and the hard socket, between the skin and a roll-on gel liner, and

between the gel liner and the socket. The negative pressure within a socket can be measured with a typical vacuum gauge in inches of mercury (inHg), where normal atmospheric pressure is 0 inHg. In discussing transfemoral suspension, the

**Table 3** (continued)

Suspension Type	Example	Primary Indication	Major Advantage	Chief Limitations
Belts (Typically used as secondary or auxiliary suspension; primary suspension only when negative atmospheric pressure cannot be used)	Silesian 	Limit socket rotation Volume changes Auxiliary suspension	Firm belt for secure feel Adjustable Can be removable	Minimal suspension Hand strength and dexterity to don Pressure around the pelvis Difficult to clean
	Elastic suspension 	Reduce socket rotation Volume changes Auxiliary suspension	Flexible Adjustable Removable	Minimal suspension Elastic suspension Hand strength and dexterity to don Warm around the pelvis Difficult to clean
	Hip joint and pelvic belt 	Short residual limbs Weak hip abductors Low levels of voluntary control	Maximum coronal support	Belt may be uncomfortable Migrates when sitting Minimal suspension Difficult to clean

larger the negative number the greater is the suspension force ( $-30$  inHg represents an absolute vacuum).

The original transfemoral suction suspension designs were used within a skin-fit socket where the suction was created simply between the skin and the inner socket wall.<sup>51</sup> Basic suction suspension systems are characterized as low, negative-pressure systems, with readings in the  $0$  to  $-8$  inHg range. During weight bearing, these systems have a vacuum reading of  $0$  inHg. During ambulation, the vacuum reading increases in value through swing phase as the momentum of the advancing limb and weight of the prosthesis attempt to distract the prosthesis from the limb.

The greater the inertial forces generated in swing, the greater these distractive forces, thus requiring higher negative pressures to hold the socket in place.

#### **Suction Suspension: Skin-Fit**

Skin-fit suction suspension has the benefit of direct skin contact with the socket, allowing high levels of proprioceptive feedback to the user. The skin moves with the socket, allowing the user to quickly perceive and react to small changes in socket position. Proximal, circumferential socket reductions create a seal against the skin that prevents air from entering the socket, thereby permitting suction to occur during swing phase. The skin

also creates surface tension along the inner socket walls that further resists some of the distraction forces felt during swing. Typically, a one-way expulsion valve is located distally on the transfemoral socket that permits air to escape during weight bearing while preventing air from entering during swing phase. For skin-fit suction suspension systems, the internal pressure clinical readings have been found to be approximately  $-8$  inHg during the swing phase.

The individual generally dons the prosthesis by initially applying a donning sleeve over the residual limb and feeding the loose end of the sleeve through the open distal valve hole. The sleeve breaks the surface tension

between the socket and the skin, allowing the individual to seat the limb inside the socket while pulling proximal soft tissue into the socket.<sup>34</sup> The sleeve is progressively and fully extracted from the socket. With the limb fully seated, the one-way air valve is installed in place.

The disadvantages of skin-fit suction suspension systems include difficulties with donning, comparatively poor mitigation of shear forces, and poor accommodation of residual limb volume fluctuations. Successful donning of a skin-fit suction suspension system requires strength and balance because the soft tissue needs to be pulled into the socket using a donning sleeve; this may be difficult for some users to manage. Scar tissue and invaginations represent another potential contraindications because shear forces are typically not well tolerated by these clinical presentations and there is potential for skin breakdown if the skin is not well protected and/or padded. Because skin-fit suction requires the maintenance of a proximal, air-tight seal against the skin, even small changes in limb size caused by a change in weight or edema can compromise suspension.

#### ***Suction Suspension: Roll-On Gel Liners and Locking Mechanisms***

Roll-on gel liners, when used as an interface, absorb shear and impact forces acting on the limb, stabilize soft tissue, and accommodate volume fluctuations. As with skin-fit suspension systems, liners are held in place by a combination of suction and surface tension and may also be used as a means of suspension with the attachment of a distal locking mechanism such as a pin, lanyard, or magnet.

To don these systems, the user rolls on the liner, inserts his or her limb into the socket, and engages a locking mechanism that is typically embedded in the distal aspect of the socket. Locking mechanisms include pins, lanyards, or

magnets. The donning of such systems is generally much quicker compared with skin-fit suction suspension systems. In addition, socks can be worn over the liner to accommodate volume changes without a loss of suspension.

Disadvantages of using roll-on liners in a transfemoral prosthesis include a minimum level of hand strength and dexterity for correct donning, the potential for tearing the somewhat fragile liners because of improper handling and sustained use, and the need for liner replacement if damage occurs. Roll-on liners also require consistently good hygiene to reduce odor and maintain cleanliness.

#### ***Roll-On Gel Liners: Hypobaric and Vacuum-Assisted Suspension***

Suction suspension with roll-on liners can be accomplished by direct contact with the liner against the socket wall (similar to skin-fit suspension) or with the use of hypobaric sealing membranes. Roll-on liners that use suction for suspension tend to have less distal distraction and minimized socket rotation compared with those that use a distal locking mechanism.

As is the case with skin-fit suction, these roll-on liner systems use negative pressure and surface tension to maintain suspension. To fully seat the residual limb and liner into the socket, the surface tension must be reduced, typically with the use of isopropyl alcohol in lieu of a donning sleeve. The liner is rolled on over the residual limb, alcohol is sprayed on the liner, and the limb and liner are slipped into the socket and engage against the inner socket wall as the alcohol quickly evaporates. The resultant seal maintains the pressure differential within the socket.

This variation in roll-on liner use can be incorporated with either simple suction or vacuum-assisted suspension. The two suspension methods differ in the internal socket pressure while standing. In simple suction suspension, the

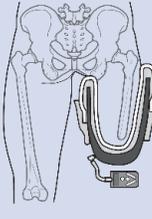
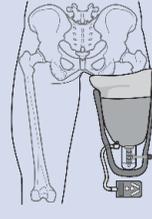
internal socket pressure is 0 inHg, whereas socket pressure with vacuum-assisted suspension is less than 0 inHg, and it can be as low as -25 inHg. Both systems use an expulsion valve to maintain the pressure differential, with vacuum-assisted suspension also using an external mechanism to draw air from the socket. Although suction systems have a negative pressure environment in swing only, vacuum-assisted systems have a continual negative pressure environment through stance and swing.

Vacuum-assisted suspension has been slow to gain acceptance in transfemoral applications.<sup>52</sup> This may be the result of complicated fabrication and donning processes and difficulties in maintaining a proximal vacuum seal. However, modern material advances, creative techniques, and design variations are making vacuum-assisted suspension a more viable choice for transfemoral applications. Anecdotal reports indicate that these systems work well for individuals with longer residual limbs and high voluntary control.<sup>49</sup>

The three basic subsichial socket designs that incorporate vacuum-assisted suspension are the single-wall internal sealing system, the single-wall external sealing system, and the double-wall internal sealing system (**Table 4**). Although these three designs each have advantages and limitations, all use a roll-on gel liner as an interface and utilize the creation of high levels of vacuum-assisted suspension between the liner and the socket rather than the skin. Although skin is compatible with basic suction suspension, it is porous and irregular in shape, making it a poor surface for maintaining elevated levels of negative pressure. Gel liners have a smooth, flexible, nonporous surface that allows for a vacuum seal to be maintained throughout ambulation, while sitting, and during participation in activities of daily living.

All three subsichial socket design systems also use a wick in the form of a

**Table 4** Transfemoral Socket Design, Subischial Variations, Vacuum-assisted Suspension

Subischial Variation	Example	Major Advantages	Chief Limitations
Subischial, single-wall system, internal sealing system (Roll-on gel liner with a hypobaric seal inside of socket, wick below the seal, external vacuum pump)		Reduce bulk when compared with other subischial designs Minimal donning process More proximal trim lines may enhance coronal stability	Limited area to achieve suspension because of the height of the vacuum seal inside of the socket Shorter residual limbs
Subischial, single-wall system, outer sealing sleeve (Long roll-on gel liner, wicking sock, reflected liner over brim of socket, seal on outside of socket, external vacuum pump)		Suspension over the entire socket Reduced trim lines Option for shorter limbs	Bulky Liner susceptible to tears at brim Lengthy donning process Exposed liner may adhere to clothing
Subischial, double-wall system, sealed internal socket (Roll-on gel liner and wicking sock, rigid internal socket affixed to liner by a sealing sleeve, internal socket connects to outer frame, external vacuum pump)		Inner suspension seal is protected by outer frame More proximal trim lines may enhance coronal stability	Bulky Lengthy donning process Lengthy fabrication and fitting process Short residual limbs

sock or fabric liner cover. The wick begins at the distal aspect of the liner and terminates distal to the vacuum seal, allowing for the transfer of air molecules between the socket and the liner. This facilitates uniform internal socket pressures between  $-5$  inHg and  $-25$  inHg; typical prosthesis users prefer approximately  $-15$  inHg. The greater the negative pressure, the greater the suspension force; however, high vacuum levels may be difficult to maintain over time.

The single-wall internal sealing design works with a hypobaric seal that is either integrated into the roll-on liner or applied over the liner. A wick is used distal to the seal, and air is drawn out of the system either manually with a hand pump or actively by with an electronic or integrated weight-activated pump.

This suspension system is simple in design, fabrication, and donning and also allows for any variation of proximal trim lines. Its main limitation is the limited surface area over which a vacuum seal can be achieved, which can be an issue for individuals with shorter residual limbs.

The single-wall, external sealing design overcomes this issue with the use of a longer roll-on liner that is reflected over the proximal brim and sealed distally on the outside of the socket using a sealing sleeve. This design requires that the proximal trim lines be reduced to allow the liner to be reflected. Although the system has the benefit of lower trim lines, the exposed liner is subject to wear from the environment and is prone to failure because of the

formation of holes. It also is bulkier than the internal sealing system.

The double-wall socket uses an internal socket to create vacuum suspension and an external socket to provide the proximal brim and distal attachment of the prosthesis. A roll-on liner is donned, followed by a wicking sock. The internal hard socket, typically half the length of the residual limb, is then applied over the liner. A vacuum seal is created when a sealing sleeve is applied over both the outer wall of the internal socket and the liner, as is often seen in transtibial systems. Negative pressures are drawn between the liner and the socket through a one-way expulsion valve. The internal socket is then inserted into the external socket where it is affixed by various forms of locking mechanisms. The

major limitations of this design are its bulk, weight, and complicated donning and fabrication processes.

### **Belt-Type or Auxiliary Suspension**

Belt-type suspension offers convenience over performance. These systems are easy to don but offer minimal primary suspension. Belt-type suspension systems are primarily used to provide secondary or auxiliary suspension and aid in control of the device. The three main types of belt-type suspension systems are the Silesian belt, elastic suspension, and the hip joint and pelvic belt (**Table 3**). Silesian and elastic systems are soft belt systems that can be attached to the socket to help reduce rotation and provide minimal suspension. For individuals who ambulate at a minimal cadence and require a socket system that is both easy to don and allows for free movement of air, these suspension systems may be an adequate primary suspension option. Alternatively, for individuals who require greater coronal stabilization because of a short residual limb or lack of abductor muscle control, a hip joint and pelvic belt can be used. This system provides maximal stability against lateral socket motion but provides minimal suspension.

## **Component Considerations**

### **Prosthetic Knee Considerations**

Because an individual with a transfemoral prosthesis has no direct musculoskeletal connection to the prosthetic knee or foot, the most optimal components must be selected. If the prosthetic knee unit is to simulate the function of the anatomic knee it must provide stability in early stance, allow for shock absorption while maintaining a lowered center of mass through midstance, provide stability through terminal stance, allow a smooth transition into swing phase, limit initial swing phase flexion across a range of cadences, advance the limb through midswing, and smoothly decelerate at terminal swing.<sup>22</sup>

The first concern in knee component selection should be stability in early stance. If the individual has limited voluntary control, the knee unit must have inherent stability, which can be variously achieved through mechanical linkages, breaking mechanisms, or hydraulic dampening control. Recently, the addition of sensors and microprocessor control units has demonstrated an increased ability to allow safe ambulation, reduced cognitive dedication to controlling the knee unit, increased gait efficiency, and increased overall user confidence with the prosthesis.<sup>53</sup>

Another concern in knee component selection is the ability of the knee to transition from stance to swing phase. The methodology varies by which the prosthetic knee “knows” when to transition from stable load bearing in stance phase to less restricted motion that will allow swing phase flexion. Mechanically controlled knee units typically rely on a transfer of load or the mechanical knee angle to initiate this transition. In contrast, microprocessor-controlled knee units use algorithms based on input received from load sensors, accelerometers, gyroscopes, and joint angles. Because of this nuanced level of regulation, microprocessor-controlled knee units allow for more controlled prosthetic ambulation, enabling the user to confidently address changes in the environment, such as walking down slopes or ramps, movements in confined spaces, descending or ascending stairs, and walking backward. These situations illustrate scenarios in which mechanically controlled knees often fail to provide consistent support and which require the prosthesis user to be cognizant of environmental changes.

### **Prosthetic Foot Considerations**

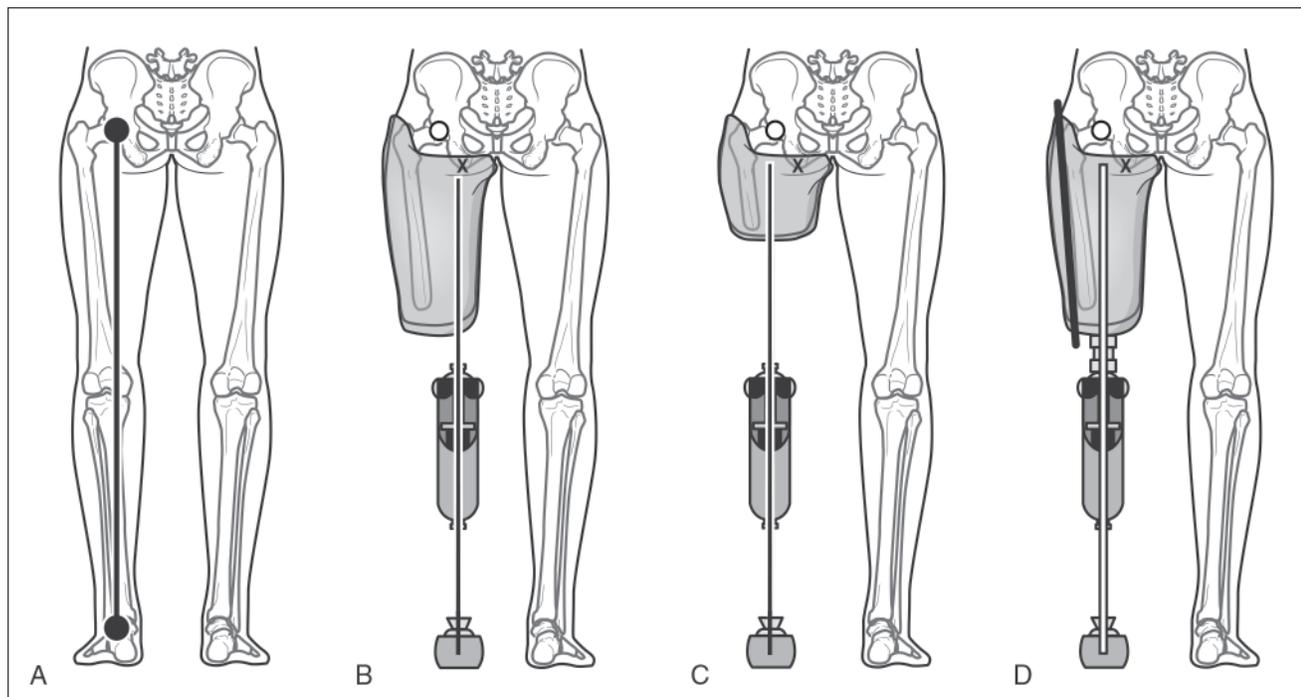
When selecting a prosthetic foot for an individual with a transfemoral amputation, an initial concern is the influence of the foot on the knee flexion moment

in early stance. If the individual has limited voluntary control, the prosthetic foot should reduce the knee flexion moment. This can be accomplished with a soft heel component in the prosthetic foot itself or by altering the alignment of the foot relative to the socket. The next concern is the transition from stance to swing phase where the foot should generally enhance late stance stability to allow the user to take an adequate step with the contralateral limb. The length, stiffness, and design of the keel, along with alternations in alignment will affect stability in this late stance phase.

### **Additional Component Considerations**

With the anatomic knee and foot absent, users of transfemoral devices are missing elements of rotation, shock absorption, and stance-phase knee flexion. Additional components of transfemoral prostheses can address these missing anatomic elements. Positional rotation units allow the prosthesis user to spin the prosthetic components distal to an adaptor. The function of these units is optimized when they are applied on the distal aspect of the socket and proximal to the knee. This allows the user to cross his or her legs, more easily tie shoes, don pants, and enter the front seat of a car. Torque absorption units can be combined with shock absorbers to reduce the shear and impact forces felt on the residual limb. Although these components are effective, they add weight, cost, and spatial considerations to the overall design of the prosthesis.

Stance flexion, the 15° to 20° of knee flexion necessary for optimized gait, is achieved during loading response,<sup>22</sup> and it is often a desired feature when creating transfemoral prosthetic devices. However, this amount of knee flexion in early stance can induce a sensation of instability for many individuals with a transfemoral amputation. Prosthetic stance flexion is variously obtained through the design of the knee frame,



**Figure 3** Illustrations show the process of initial coronal alignment for placement of the prosthetic knee and foot. **A**, Posterior view of anatomic alignment in the coronal plane. The hip joint aligns over the knee joint and the ankle joint. **B**, Because the hip joint is difficult to represent in transfemoral alignment, the location of the ischium (X) is used instead. For longer residual limbs, the coronal alignment line begins 1 inch (2.54 cm) lateral to the ischium, then continues distally through the posterior bisection of the prosthetic knee and the posterior bisection of the prosthetic foot. **C**, For shorter residual limbs or individuals with lower levels of voluntary control, the coronal alignment line is shifted laterally, closer to the bisection of the prosthetic socket, but never more lateral than the bisection of the socket. **D**, A completed transfemoral alignment with the socket set at the initial adduction angle (heavy black line) and the connecting pylons attached to the socket, knee, and foot.

compressive bumpers, and hydraulic cylinders.

### Alignment Considerations

The principles of transfemoral prosthesis alignment have altered little over time. In 1955 Radcliffe<sup>14</sup> addressed transfemoral alignment considerations in his statement that the artificial limb "... must provide both adequate support and a natural-appearing gait with as modest consumption of energy as possible." These standards have not appreciably changed. Prosthetists attempt to create a stable and effective transfemoral gait pattern with proper socket fit; effective suspension; and diligence in bench, static, and dynamic alignments.

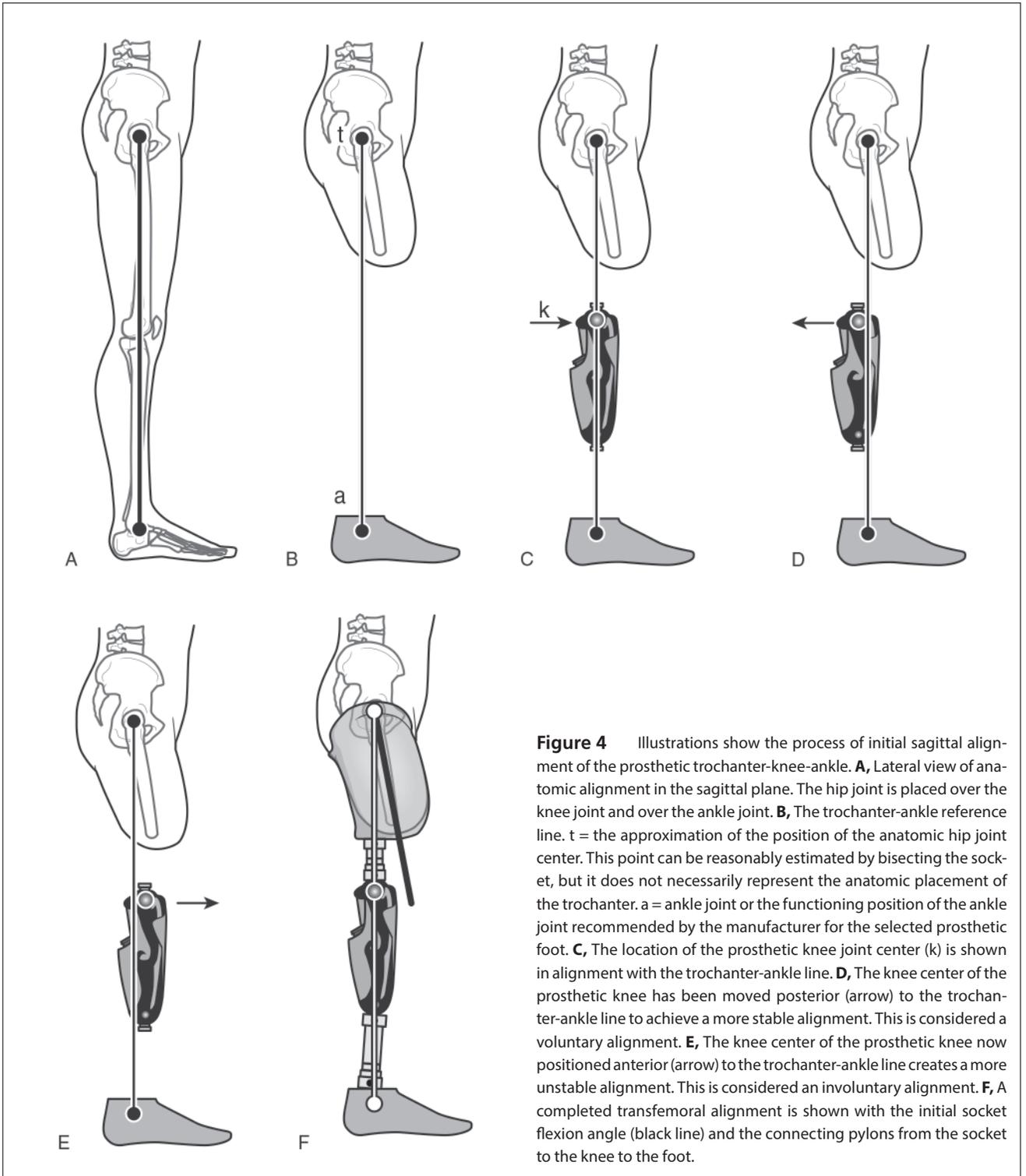
Before it is fitted to a patient, the prosthesis is set up in bench alignment, which reflects the individual's

hip flexion, adduction attitude, and transverse limb orientation. The socket is generally set in a flexion angle 5° greater than the individual's maximum hip extension. This added hip flexion permits the user to take an adequate step with the contralateral limb and puts a mild stretch on the hip extensor muscles to allow them to be more efficient in early stance.<sup>14,29,30,37,40</sup>

The transfemoral socket is also set to match the individual's recorded adduction orientation. This will align the femur under the hip joint and put a mild stretch on the gluteus medius, increasing efficiency during single-limb support.<sup>54</sup> Setting the proper amount of socket adduction also reduces the tendency for proximal lateral gapping of the socket and helps to maintain a narrow base of support.<sup>29,40</sup> Transverse orientation is determined by the user's line of

progression and the necessity to minimize transverse plane gait deviations. This orientation is especially important for proper fitting of IC sockets.

With the socket in the proper orientation, the focus is on the placement of the prosthetic knee and foot. In able-bodied individuals, coronal alignment of the hip joint is typically directly over the knee and ankle joints (**Figure 3, A**). For initial bench alignment of the transfemoral prosthesis, the actual hip joint cannot be used as a reference point because it cannot be located on the prosthetic socket. However, locating a point on the socket brim that is 1 inch (2.54 cm) lateral to the location of the ischium will provide a reasonable approximation. The prosthetic knee and ankle joints are placed directly below this identified point. The initial coronal bench alignment allows for stability in



**Figure 4** Illustrations show the process of initial sagittal alignment of the prosthetic trochanter-knee-ankle. **A**, Lateral view of anatomic alignment in the sagittal plane. The hip joint is placed over the knee joint and over the ankle joint. **B**, The trochanter-ankle reference line.  $t$  = the approximation of the position of the anatomic hip joint center. This point can be reasonably estimated by bisecting the socket, but it does not necessarily represent the anatomic placement of the trochanter.  $a$  = ankle joint or the functioning position of the ankle joint recommended by the manufacturer for the selected prosthetic foot. **C**, The location of the prosthetic knee joint center ( $k$ ) is shown in alignment with the trochanter-ankle line. **D**, The knee center of the prosthetic knee has been moved posterior (arrow) to the trochanter-ankle line to achieve a more stable alignment. This is considered a voluntary alignment. **E**, The knee center of the prosthetic knee now positioned anterior (arrow) to the trochanter-ankle line creates a more unstable alignment. This is considered an involuntary alignment. **F**, A completed transfemoral alignment is shown with the initial socket flexion angle (black line) and the connecting pylons from the socket to the knee to the foot.

double-limb stance, induces a modest lateral thrust in single-limb stance, and achieves a narrow 2-inch (5.08-cm) base of support (**Figure 3, B**). The prosthetic knee and ankle should be

placed more laterally under the socket for shorter residual limbs or in individuals with compromised voluntary control (**Figure 3, C**). However, this necessary accommodation will increase energy

expenditure by inducing a wider base of support.

In able-bodied individuals, the sagittal plane alignment of the ground reactive force is posterior to the hip joint and

anterior to the knee and ankle (**Figure 4, A**). This anatomic alignment allows the prosthesis user to stand with minimal energy expenditure. Because the anatomic hip joint cannot be used as a point of reference on the prosthetic socket, the apex of the greater trochanter is used. For bench sagittal plane alignment, a simulated reference line is used to create a stable prosthetic alignment. This line is called the trochanter-knee-ankle line. To understand the use of the trochanter-knee-ankle reference line, a single axis knee and single axis foot will be assumed, because these components have very little inherent stability and require that the stability of the overall prosthetic system be derived by the alignment of the socket relative to the knee and ankle components.

The trochanter-knee-ankle line begins by determining a reference point for the trochanter and approximating the position of the hip joint. This point can be reasonably estimated by bisecting the socket in the sagittal plane at its most proximal aspect (**Figure 4, B**). This is followed by the placement of the prosthetic ankle joint, a reference point that differs for every prosthetic foot and is identified within individual manufacturer's guidelines. When the trochanter and prosthetic and ankle reference points are vertically aligned (**Figure 4, B**), the prosthetic knee is set at its proper height and located according to the manufacturer's recommendation for the knee center's sagittal reference point. This point may be posterior, through, or anterior to the trochanter-ankle line (**Figure 4, C**). Placing knee center posterior to the line creates a safe alignment, because the individual's weight and the ground reaction force keep the knee locked in extension. This also can be described as an involuntary alignment, because no voluntary control is required to keep the knee in extension (**Figure 4, D**). In contrast, placing knee center anterior to the trochanter-ankle line creates an alignment in which the individual

has to voluntarily control the sagittal stability of the knee. This alignment is also seen with knee units possessing inherent stability because it facilitates early stance flexion and permits easier initiation of swing phase knee flexion in late stance (**Figure 4, E**). If the knee center point is directly on the trochanter-ankle line, it is considered to be on "trigger," where the system may be in voluntary or involuntary alignment depending on the placement of the prosthetic foot with each step (**Figure 4, F**).

In the transverse plane, the prosthetic knee is externally rotated 5° to compensate for the natural 5° of internal socket rotation that will occur during swing phase. This rotation ensures that the knee will flex in the line of progression during swing. For individuals who walk faster, the amount of internal rotation will increase, and the initial external knee rotation should be larger.

When bench alignment is complete, the prosthesis is donned and static or standing alignment is assessed. The foot should be flat on the floor with 2 to 4 inches (5.08 to 10.16 cm) of base support. The knee should be extended and safe with socket flexion, with adduction and rotation matching the individual's limb orientation. Generally, if bench alignment conditions were observed, minimal adjustments will need to be made to achieve a proper static fit. Any accommodative changes in flexion or adduction will change the position of the socket over the knee and foot, which will alter the stability of the system. In such instances, the proper trochanter-knee-ankle alignment should be reestablished before ambulation begins.

Because every prosthetic knee and foot has different triggers to transition from stance to swing control, the prosthesis user must be made aware of how each component functions before ambulation is attempted. The prosthesis user should be observed and instructed on proper techniques while functions such as sitting, bending the knee, and

advancing the limb are practiced in a safe environment (such as with parallel bar support before ambulation). During dynamic alignment, the prosthetist will work to optimize gait and minimize energy expenditure by making incremental changes to the alignment and working with a therapist to focus on enhancing muscle strength and range of motion.

## Summary

Great technological advances have been made in transfemoral prosthetic sockets, components, and suspension systems in recent years. However, it must be understood that there is no single socket design, alignment, or prosthetic system that will be optimal for all individuals with transfemoral amputation. There is a need for prosthetists to continually develop their clinical and technical skills to provide the most appropriate device and the best fit for each patient.

The team must address issues of energy expenditure, body image, levels of voluntary control, and socket fit in creating the treatment plan, and they should have an intimate knowledge of appropriate socket designs, suspension systems, components, and alignment considerations. Clinical experience, knowledge of evolving clinical standards, the use of available evidence, and the incorporation of appropriate outcome measures also will assist the rehabilitation team in providing optimal care to their patients.

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