Effect of Variance in Prosthetic Foot Stiffness on Muscle Function and Sound Limb Mechanics

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Clinical Question: Does variance in prosthetic foot stiffness affect muscle function and sound limb mechanics?

Background: The importance of choosing the appropriate prosthetic foot for a person with limb loss is recognized, but the effects of foot choice on patient presentation and the considerations for clinical outcome are not well defined. Today's prosthetic feet store and return energy, affecting the body's reliance on residual musculature for support and propulsion.[1] Altering the stiffness of the prosthetic keel will affect energy costs, muscle activity, muscular compensation, gait mechanics, and strategy.[2-4] A better understanding of how prosthetic foot mechanical properties effect patient outcomes is needed.

Search Strategy:
Databases Searched: Google Scholar, PubMed
Search Terms: transtibial, amputee, prosthesis, rigidity, stiffness, keel, foot
Eligibility Criteria: Articles published in English between 2010-present with focus on effects of prosthetic foot componentry in unilateral transtibial amputees.

Synthesis of Results: Fey et al 2011 reported decreasing keel stiffness, decreased mechanical efficiency (energy storage and return), increased reliance on musculature for body support, and decreased muscle contributions to propulsion and swing initiation. Fey et al's 2012 article on optimizing prosthetic foot stiffness[2] further identified the influence of altering prosthetic keel stiffness on muscle and foot function by defining the anatomical compensations and mechanical responses. Contradicting previous findings, modeling analysis revealed that throughout residual limb stance as keel stiffness decreased, keel contribution to body support and breaking forces increased. The primary drivers of forward propulsion and swing initiation through stance became the hamstrings (knee flexors) and gravity. In early stance, the hamstrings facilitated propulsion. In late stance, the rectus femoris transferred power from the residual limb to the trunk. Through dynamic coupling of leg and trunk, late stance rectus femoris activity functioned as an important mechanism facilitating propulsion through forward momentum. Fey et al's 2012 article on altering prosthetic foot stiffness[3] expanded on the author's optimizing prosthetic foot stiffness article by generating design-optimized prosthetic feet coupled with dynamic simulations of amputee walking to identify the optimal foot stiffness that minimizes metabolic cost and intact knee joint loading. Fey et al used simulation analysis to optimize foot design on a patient-specific basis. The authors discovered that decreasing keel and ankle stiffness significantly reduced metabolic cost but increased sound leg loading. Increasing keel stiffness (maintaining ankle stiffness and decreasing heel stiffness) significantly decreased sound leg loading but increased metabolic cost.

Clinical Message: The choice of prosthetic foot significantly impacts walking strategy, gait mechanics, and performance.[2-5] Understanding how the body responds to various levels of prosthetic foot stiffness enables the practitioner to make informed clinical decisions. Muscular contributions to support propulsion and the inverse relationship between metabolic cost and sound leg loading are greatly affected with altered keel stiffness. As keel stiffness decreases, sound leg loading increases and metabolic cost decreases (due to decreased reliance on musculature for support).[2,3] As keel stiffness increases, metabolic cost increases and sound leg loading decreases. Reliance on residual limb and core musculature is increased to control the mechanical breaking forces that provide overall support and stabilization; which in turn decreases reliance on the sound limb and increases metabolic cost.[2,3] Prosthetic foot design, componentry additions, foot shell, and shoe wear all affect prosthesis function.[1-6] Further research is needed to understand their effects on amputee gait. Through a better understanding of the effects mechanical properties have on sound and residual limb mechanics, the prosthetic foot prescription can be customized to address deficiency, limit acquisition of poor gait strategies, reduce overreliance on the sound limb, and address muscular dysfunction and compensation.

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Evidence Table

	Fey, 2012[3]	Fey, 2012[5]	Fey, 2011[4]
Population	Unilateral transtibial amputees	Unilateral transtibial amputees	Twelve male unilateral transtibial amputees (N=12), (mean age= 51.9) (std. dev. = 17.1). Mean post-amputation time = 13.9 yr. with std. dev. = 14.4
Study Design	A forward-dynamics driven modeling and simulation analysis	A forward-dynamics driven modeling and simulation analysis	Crossover study design (study design in which subject receive a series of different treatments, which are then compared)
Purpose	To identify the influence of prosthetic foot stiffness on muscle and foot function by developing forward-dynamics simulations of below-knee amputee walking with a range of foot stiffness levels	Couple design optimization of energy storage and return (ESAR) prosthetic feet with forward-dynamics simulations of amputees walking to identify the optimal foot design that improves metabolic cost and joint loading during below-knee amputee walking	To identify the influence of energy storage and return (ESAR) foot stiffness on gait characteristics by manufacturing selective laser sintering (SLS) ESAR feet with a range of stiffness levels and quantifying their effect on below-knee amputee walking. Specifically, to investigate the influence of ESAR foot stiffness on gait kinematics, kinetics, muscle activity, prosthetic ESAR, and mechanical efficiency during over-ground walking

Intervention	Applied varying prosthetic foot stiffness to a forward-dynamics driven musculoskeletal model of unilateral transtibial amputee walking using a planar bipedal musculoskeletal model	Manipulated parameters applied to muscle-actuated forward-dynamics driven simulations of unilateral transtibial amputee walking to identify the optimal prosthetic foot design that minimizes the biomechanical quantities of metabolic cost and intact leg loading	Twelve transtibial amputees walked across 12 meters of 4 force plates using three different prosthetic feet of varying rigidity and in a randomized order. A certified prosthetist orthotist (CPO) confirmed proper prosthetic fit and alignment. At least five force-plate contacts per leg were measured per condition
Comparison	Three musculotendon actuated models walking with a range of varying prosthetic foot stiffness levels: stiff, nominal, and compliant	Differences in muscle and prosthetic foot function between simulations	Comparison between sound side and prosthetic side under three different conditions
Methodology	Researchers generated three 2D muscle-actuated forward dynamic simulations of unilateral transtibial amputee walking with a range of prosthetic foot stiffness levels. Experimental variables were assessed Prosthetic foot models were based off of Freedom Innovations Highlander prosthetic foot. The model generated consisted of 22 rigid segments connected in series. Eighteen viscoelastic elements at each degree of freedom were used to model foot stiffness	Researchers generated a planar bipedal musculoskeletal model of a left unilateral transtibial amputee walking using SIMM. Muscle-actuated forward dynamic-driven simulations were run with a range of prosthetic foot stiffness levels applied Prosthetic foot models were based off of Freedom Innovations Highlander prosthetic foot. The model generated consisted of 22 rigid segments connected in series. Eighteen viscoelastic elements at each degree of freedom were used to model foot stiffness	Biomechanical analysis of 12 unilateral transtibial amputees walking over ground (>= 1.2 m/s) with three different prosthetic feet of varying keel and heel stiffness (feet were manufactured using additive manufacturing)

Outcomes	Muscle and foot contributions to body support (vertical GRF), propulsion (A/P GRF) and residual leg swing (residual-limb mechanical power) using GRF decomposition, and segmental power were tested	Calculated individual muscle and foot contributions to body support (vertical GRF), forward propulsion, residual-limb swing, metabolic cost (MetE), and IL knee contact force (JCont)	Differences in peak GRFs, sagittal plane joint angles and moments, EMG magnitudes, and prosthetic energy quantities
Key Findings	As stiffness decreased: First half of stance: · Prosthetic keel provided increased support (neg. propulsion) · Heel provided decreased support · Hamstrings provided decreased support and increased propulsion Second half of stance: · Prosthetic keel provided decreased propulsion and required increased support · Prosthetic keel absorbed less power from leg (decreased swing initiation) requiring muscle compensation · Sound limb vastus provided increased support, and residual limb rectus femoris energy expenditure increased from leg to trunk to aid propulsion	As stiffness decreased: · Metabolic cost decreased · Sound Limb loading increased o Increased keel support As stiffness increased: · Metabolic cost increased · Sound limb loading decreased	As stiffness decreased: · Peak residual-limb and sound-limb ankle angles increased · Residual-limb plantar flexion moment and second peak GRF decreased · Residual-limb changes in joint kinematics were most apparent in single-limb support and terminal double limb support and consistent with a flexed body posture (decreasing support and increasing energy expenditure) · Residual-limb knee flexion angle increased · Residual-limb peak moments became extensor centric from mid to late stance o Increased vastus activity second half of stance o Decreased biceps femoris (hamstring) activity second half of stance ▪ To initiate swing

			• Residual-limb glute med. activity increased throughout stance o Residual-limb increased glute med. and vastus activity were consistent with an increased role in providing body support • Residual-limb and sound-limb breaking GRF increased • Residual-limb and sound-limb early stance knee ext. moments (K1) increased o Providing additional support • Sound-limb vastus and rectus femoris activity increased (providing support) • Sound-limb early stance hip extensor moments increased to provide breaking and body support during first half of stance

Study Limitations	1. No patient intervention 2. Parameters were estimated from previously collected in-vivo trials	1. Influence of the different stiffness profiles on muscle and foot function need to be experimentally verified 2. It is not clear if the biomechanical objectives optimized in the study can be utilized by the central nervous system (CNS) to optimize walking in a similar manner 3. Equal objective costs were assumed when reducing both metabolic cost and intact knee contact forces 4. Parameters were estimated from previously collected in-vivo trials	1. Feet were manufactured using additive manufacturing; study authors did not use pre-existing prosthetic feet 2. Assumptions were made when defining prosthetic foot segments and residual-limb ankle joints for inverse dynamics 3. The same prosthetic feet were used on subjects of varying weight 4. Only male patients participated 5. Only transtibial amputees

Abbreviations:

1. ESAR = Energy Storage and Return
2. SLS = Selective Laser Sintering
3. GRF = Ground reaction force
4. PF = Plantar Flex
5. Glute Med. = Gluteus Medius
6. Vastus = Vastus Lateralis