Osteoarthritis in Amputees: A Double Whammy?

Prevalence of Osteoarthritis in Amputees

Currently, 1.7 million individuals in the United States have had an amputation with another 185,000 occurring each year. [1,2] Lower extremity amputees face a number of musculoskeletal issues secondary to amputation that can increase disability. Osteoarthritis is common musculoskeletal issue that arises in lower extremity amputees with long-term prosthetic use. [3] Vincent et al report, "osteoarthritis (OA) is the most frequent cause of disability in the United States." [4] Studies show osteoarthritis is more prevalent in lower extremity amputees than the general population. [5-7] Such limitations often lead to poor mobility and in turn decreased participation and quality of life. [8]

Osteoarthritis (OA), a common condition with aging, is a chronic degeneration of articular cartilage and periarticular bone remodeling that causes joint pain and stiffness and results in functional impairment. [8] Morgenroth et al state, "the etiology of OA is thought to be multifactorial: a combination of potentially modifiable factors related to abnormal joint mechanics, superimposed on underlying risk factors including age, gender, race/ethnicity, and other specific genetic factors." [8] Amputees are more likely to have OA in their intact limb compared to the amputated limb and transfemoral amputees are at a greater risk than transtibial amputees. [8,9] Struyf et al determined knee and hip OA was more prevalent in the intact limb of amputees than in the general population. [5] The prevalence of knee OA in amputees was reported to be 28.3% for men and 22.2% for women whereas the prevalence in able-bodied men was 1.58% and 1.33% for women. The prevalence of hip OA was also greater in male and female amputees, 15.3% and 11.1% respectively, compared to 1.13% of able-bodied men and 0.98% of able-bodied women. [5]

Contributors to OA in Amputees

Amputees are at a greater risk for developing OA from altered gait mechanics with prosthetic use. A majority of amputees who use prostheses for ambulation do so with at least one gait deviation, either from poor prosthetic fit, poor gait training, or bad habits. [3] Amputees often ambulate with an increased demand on the intact limb from temporal asymmetries, asymmetrical muscle activity, and increased joint loading with compensations, all of which can result in degenerative changes at the hip and/or knee.

<u>Temporal Asymmetries</u>

Numerous studies have shown that amputees have greater stance time and shorter swing time of their intact limb. [3,10-12] Sanderson et al reported amputees had statistically greater stance times at 1.2 m/s and 1.6 m/s on the intact limb (798ms and 667ms) compared to the prosthetic limb (758ms and 624 ms). [10] Further, at both 1.2 m/s and 1.6 m/s, the intact limb had shorter swing times (427 ms and 394ms) compared to the prosthetic limb (468 ms and 410 ms). [10] When compared to able-bodied subjects, amputees had greater stance and shorter swing times on the intact limb indicating asymmetries and a greater demand on the intact limb than in normal walking. [10] Dingwell et al also reported significant differences in temporal asymmetries between amputees and able-bodied subjects with percent stance time, push off, and single support time being 4.6 times greater for amputees compared to able-bodied subjects. (Figures 1 and 2) [11] Kovac et al also reported significant differences when comparing amputee gait to able-bodied gait with amputees' prosthetic limb having increased swing time and decrease stance time. [12]

Sixty-four percent of amputees report more dependence on their intact lower extremity than amputated extremity during activities. [3] This is, in part, shown by the increase in stance time on the intact lower extremity. With an increase in time spent on the intact limb comes an increase in demand to the hip and knee on that side. Burke et al determined the increase in stance time seen in the intact limb could be one contributor to an increase in prevalence of hip and knee OA in lower extremity amputees. [6]

Strength Asymmetries

Amputees' preference of the intact limb over the prosthetic limb further exacerbates articular cartilage degeneration through muscle strength asymmetries. Amputees commonly have atrophy of muscles on the residual limb side and hypertrophy of muscles on the intact side from increased use of the intact limb and lack of use in the prosthetic limb. [13] An increase in muscle activity on the intact limb will increase the contact force and pressure at the joints of intact limb, contributing to OA. The increase in intact limb dependence could be attributed to lack of confidence in the prosthesis or decreased proprioception in the residual limb. [14] Isakov et al report statistically significant differences were seen in hamstring and quadriceps strength measured concentrically, eccentrically, and isometrically with the intact limb being significantly stronger. [15] Nadollek et al found a correlation between increased weight bearing on the residual limb and stronger hip abductors indicating improved strength symmetry for improved stance symmetry. [14] Lloyd et al also examined strength asymmetry for correlations. [13] Knee external adduction moment loading rate (KEAMlr) strongly correlated with knee extension symmetry angle and a moderately with knee flexion symmetry angle. [13] A strong correlation was also found between knee flexion symmetry angle and the ground reaction force-loading rate of the intact limb. [13] Lloyd et al found strength asymmetries and determined they contributed to OA through asymmetrical loading rates taxing the intact limb. (Figure 3)[13]

In another study, Sadeghi et al discuss the intact limb's role in gait compensations. [16] The research group determined that amputees ambulate with significantly higher power bursts of the hip musculature on the intact limb at midstance and push off. [16] Further, an increase in hip extension was seen at heel contact. Sadeghi et al explained this as a compensation for the absence of ankle plantarflexion and forefoot movement at push off in that the hip extensors on the intact limb are propelling the body forward from behind as well as for trunk control. [16] At push off, the intact limb demonstrated increased plantarflexion activity, decreased hip flexion activity and increased knee extension activity. [16] During double support, heel contact of the intact limb occurs while the dorsiflexors and knee extensors are acting as shock absorbers during the transfer of body weight while the hip extensors are providing the force to propel the body. [16] Lloyd et al discuss this gait pattern stating, "the intact limb must produce the force necessary to propel the body over the prosthetic limb, and absorb the impact as the center of gravity falls back down...this 'moment avoidance' gait strategy therefore results in increased load rates and muscular demands on the intact limb, while sparing the prosthetic side limb." [13] Protecting the residual limb from large moments results in high internal and external moments in the intact limb. Lloyd et al conclude, "large strength discrepancies between sides may therefore be related to decreased function of the prosthetic side limb, asymmetrical limb loading and an increased OA risk in the intact limb."[13]

<u>Abnormal Joint Loading</u>

External knee adduction moments and internal knee abduction moments have been shown to correlate with severity of OA in abnormal joint loading. [8,17] In normal ambulation, the ground reaction force lies medially to the knee joint axis creating an external adduction moment. An internal abduction moment is needed to counter the external adduction moment for successful stance.[8] Consequently, an increase in external adduction moment from poor prosthetic alignment or gait deviations results in an increase in internal abduction moment from the musculature, which in turn increases joint compression of the intact limb.[8] Norvel et al state, "the gait abnormalities exhibited by amputees with a prosthesis may result in abnormal joint loading that, over time, may lead to joint pain and degeneration" in the intact limb[9]

Royer et al examined hip and knee frontal plane moments in amputees as it has been shown to relate to OA in able-bodied subjects. A significant difference between the peak internal abduction moment of the intact limb and residual limb was seen for both the hip and knee. The intact internal knee abduction moment was 0.55 ± 0.18 N m/kg whereas the prosthetic side was 0.38 ± 0.22 N m/kg ($p=0.028$).[17] The intact internal hip abduction moment was 0.88 ± 0.22N m/kg on the intact limb and 0.63 ± 0.19 N m/kg on the prosthetic side ($p=0.01$). (Figure 5)[17] The internal knee and hip abduction moments were 46% and 39% greater, respectively, than the prosthetic side and 17% and 6% greater, respectively, than normal values.[17] Further, the prosthetic side had 23% smaller internal knee abduction moment and 31% smaller internal hip abduction moment compared to normal values.[17] Greater moments in the intact limb compared to the prosthetic limb and normal values indicate a greater risk of joint degeneration in both the knee and hip with the knee being at a greater risk than the hip.

Residual Limb Length and Amputation Level

The length of the residual limb or amputation level contributes to osteoarthritis in the intact limb through associated gait deviations. Studies show a shorter residual limb is associated increased pelvis and trunk excursions.[18,19] Further, differences in joint moments and power generation with gait are seen between transtibial and transfemoral amputees when compared to able-bodied subjects.[20]

Bell et al compared the gait of amputees with short residual limbs (21% to 56% of the intact limb length) to long residual limbs (57% to 77% of the intact limb length).[18] The research group reported statistical significance in gait speed, differences at the trunk, differences at the pelvis, and prosthetic abduction.[18] Amputees in the short residual group ambulated at a slower self-selected gait speed compared to the long residual limb group, 1.22 ± 0.1 m/s and 1.37 ± 0.13 m/s respectively ($p = 0.004$).[18] Differences seen at the trunk include forward trunk flexion and lateral trunk flexion. Amputees in the short residual group had greater forward trunk flexion of 6.7° ± 1.85° whereas the long residual limb group had 4.3°±0.99° of trunk flexion, a difference of 2.3° ($p = 0.003$).[18] Lateral flexion was also significantly greater in the short residual limb group, 9.8° ± 1.74° compared to 6.7° ± 1.96°, a 3.6° difference ($p=0.001$).[18] Differences at the pelvis include pelvic tilt and pelvic obliquity. The short residual group ambulated with 11.8° ±2 .47° of pelvic tilt and 9.8° ± 3.36° or pelvic obliquity compared to 8.2° ± 2.14° or pelvic tilt and 6.9° ± 1.60° of pelvic obliquity in the long residual group, a difference of 3.6° for pelvic tilt and 2.9° of pelvic obliquity.[18] Baum et al also reported a significant correlation between residual limb ratio and pelvic tilt excursion ($R^2=0.465$).[19] Bell et al report these differences are due to loss of hip stabilization, indicating strength training of the hip musculature. Further, a significant difference was seen in hip abduction of the prosthetic limb with hip abduction in shorter residual limbs being 9.7° ± 3.46° compared to 7.1° ± 2.64° in the long residual limb group.

The results of this study indicate decrease musculature and lever arm secondary to shorter residual limb length causes gait deviations in temporospatial and kinematic outcomes.[18]

Nolan et al examined compensations of the intact limb with net joint moments and power output in transfemoral and transtibial amputees compared to able-bodied subjects at a gait speed of 1.2m/s ± 3%.[20] Compensations seen include increased ankle range of motion, increased knee and hip extensor moments, increased knee power output, and increase hip power absorption at weight acceptance. [20] Increased knee extensor and hip flexor moment, knee power absorption and hip power output compensations were seen at push off. [20] The transtibial amputees had significantly greater peak maximal knee extensor and hip flexor moments and peak knee and hip power output in the intact limb compared to their able-bodied counterparts.[20] The transfemoral group had significantly greater peak ankle dorsiflexion moments, peak knee and hip extensor moments, and peak power output at the knee. They also had greater peak power at the hip but it was not significant. (Table 1, 2) [20] These gait deviations, increases in power output over the intact limb and joint moments contribute to joint degradation through increased demands via contact force and contact area.

Prosthetic Components

Prostheses components also contribute to OA in that different types work to decrease the load through the intact limb. Morgenroth et al, Grabowski et al, and Hill et al examined the effects of prosthetic foot type on the loading of the intact limb. [21-23] An increase in loading of the intact limb results from a decrease in push off of the prosthetic limb, thus improving the push off in the prosthetic limb will decrease forces through the intact limb during the double support phase.[21] Morgenroth et al explain, “if the trailing prosthetic limb produces a reduced push-off, the leading intact limb must perform a greater share of center of mass velocity redirection, thus increasing the ground reaction loading impulse on the leading limb." [21] This increase in load increases the external adduction moment at the knee, which correlates with joint degeneration.[21] Morgenroth et al compared three different prosthetic feet with different varying amounts of push-off to determine the effects of push-off on the intact limb. Prosthetic feet included a controlled energy storage and return foot (CESR), a conventional foot (Seattle LightFoot2), and the foot prescribed to the subject (dynamic elastic response). [21] The research group reported a statistically significant negative relationship between the push off of the prosthetic limb and the initial peak knee external adduction moment of the intact limb, indicating an increase in prosthetic push off decreases the external adduction moment in the intact limb.[21] The CESR foot provided 68% greater push off compared to the prescribed foot and 137% compared to the conventional foot.[21] Further, the initial external adduction peak was significantly lower in the CESR and prescribed feet when compared to the conventional feet. [21] (Figure 6)

Grabowski et al examined ambulation at different speeds with a passive-elastic foot and a powered ankle-foot compared to able-bodied counterparts. Results show the powered ankle-foot, when compared to the passive-elastic foot, decreased the peak resultant forces through the intact limb when walking at slow to moderate speeds. [22] The ground reaction forces in the intact limb were also significantly lower with the powered ankle-foot when ambulating 0.75-1.50 m/s, averaging 6.6% lower impact peak ground reaction forces. [22] The external adduction moment was significantly lower with the powered ankle-foot prosthesis compared to the passive-elastic prosthesis at 1.50 m/s and

1.75 m/s, 20.6% and 12.2% respectively.[22] There was a 5-22% decrease at speeds of 0.75 m/s, 1.00 m/s, and 1.25 m/s but these values were not found to be significant.[22] When comparing amputees to able-bodied subjects, ambulating with passive-elastic foot prosthesis produced greater peak resultant forces in the intact limb whereas the powered ankle-foot prostheses peak resultant forces were not significantly different from able-bodied values.[22] (Figure 7)

Hill et al report similar results when comparing passive ankle-foot prostheses to powered ankle-foot prostheses (BioM) at 1.25 m/s in a case series. There was a decrease of 8% in the peak resultant force (PRF), 18% decrease in force loading rate (FLR), an 8% decrease in peak heel-strike foot pressure (PP), and a 15% decrease in the initial peak knee external adduction moment (EAM) with the powered ankle-foot compared to the conventional foot.[23] There was a 49% decrease in lead leg transition work from step to step and a 334% increase in trailing leg transition work from step to step in the powered ankle-foot prosthesis compared to the passive foot.[23] These studies indicate the prosthetic foot prescription plays an important role in the loading of the intact limb and should be considered when prescribing prosthetic components as a preventative measure to OA in the intact limb.

<u>Bone Mineral Density</u>

Bone remodels based on the load or lack there of that is placed on it. Bone increases in density with an increase in load and decreases in density with a decrease in load.[24] Royer et al report, “increased bone mineral density may be associated with increased risk of OA as a decrease in bone compliance places excessive wear stress on articular cartilage.”[17] An increase in the bone mineral density in the proximal tibia with a larger internal knee abduction moment has been seen in healthy individuals.[24] In another study, Royer et al examined bone mineral density in amputees intact and prosthetic limb compared to controls.[24] Results showed the intact limb of amputees had 45% and 10% greater bone mineral density in the medial knee than the residual limb and control, respectively.[24] The bone mineral density of the femoral neck was 12% greater in the intact limb compared to the prosthetic limb but was no different than the controls.[24] Results from both studies show the hip and knee of the intact limb experiences increased loading and increased bone mineral density.[17,24] The hardening of subchondral bone, making it a poor absorber, coupled with joint loading further promotes joint degeneration.[13]

Implications of OA

The implications of OA in amputees are substantial. Joint pain and decreased mobility secondary to articular cartilage breakdown can result in long-term disability.[5] Functional impairment and decreased independence occur in older adults with knee OA being a leading cause of morbidity.[8] This is even more drastic when compounded with a previous disability, such as an amputation. Morgenroth et al state, "many individuals with a lower extremity amputation face mobility challenges at baseline … OA in the joints of the intact limb can have an additive debilitating effect on mobility and quality of life in this population."[8] Pain in the intact limb from OA can negatively impact mobility and consequently participation in vocational, educational, or social activities as OA in the intact limb negatively correlates with prosthetic use.[25] Norvell et al report, "a greater proportion of amputees than nonamputees reported that their pain kept them from their usual activities for more than 30 days."[26] This decrease in participation can in turn negatively affect quality of life of amputees. Amputees with comorbidities are more likely to be less mobile, which decreases their independence with ADLs; independence in ADLs is a strong predictor for quality of life.[25] Geertzen et al report comorbidities, like osteoarthritis, result in a significant decrease in the Reintegration to Normal Life (RNL) score.[25] Maintaining the integrity and joint health of amputees' intact limb is crucial in preventing further disability or setbacks.

Preventative measures should be taken to reduce the risk of OA in amputees immediately following amputation to preserve long-term function. Correcting temporal asymmetries, muscle strength asymmetries, and decreasing joint loading through prosthetic adjustment, gait training, and strength training will allow for a more symmetrical gait with more symmetrical joint loading. Addressing these factors that contribute to the development of OA will reduce the stresses felt in the hip and knee and thus prevent joint degeneration in the intact limb.

Prevention and Treatment

Mechanical Preventions and Treatments

Much like the mechanical treatments in the general population for OA, such as valgus bracing and lateral wedges, mechanical modifications can be made to the prosthesis to improve joint loading during gait.[8] Proper socket fit and prosthetic alignment can decrease abnormal loading of the intact limb by altering the external adduction moment and thus decreasing the internal abduction moment.[3,8] Morgenroth et al report, "increased trailing limb push off and feet that are functionally arc shaped during gait with larger radius of curvature are associated with reduced leading-limb loading."[8] The type of foot component on the prosthesis can also alter the moments in the intact limb. Amputees who ambulate with energy storage and return foot have a 13% reduction in external adduction moment in the intact limb compared to a traditional foot.[8] Significantly increasing the foot-ankle push-off of a prosthetic foot results in a 26% reduction of the initial peak external adduction moment in the intact limb.[8] Morgenroth et al state, "prosthetic feet with optimized roll-over shape…also have the potential to decrease intact limb loading."[8]

Gait Training

After amputation, the decrease in body weight, change in center of mass over the base of support, altered weight acceptance, single limb support time, and limb advancement all contribute to gait asymmetries.[27] As amputees have decreased stance time on the prosthetic limb, thereby increasing the load felt in the intact limb with reciprocal gait, gait training in physical therapy to correct gait deviations and temporal asymmetries can aid in preventing joint degeneration in the intact limb. In a study by Dignwell et al, it was confirmed that amputees have significantly greater asymmetries than healthy subjects, specifically in percent stance time, push off force, and single support time.[11] After real time visual feedback was provided, amputees were able to significantly decrease the amount of asymmetries seen in percent stance time, push off force, and single support time.[11] Results indicate amputees can correct their gait mechanics with feedback to a more symmetrical gait pattern and consequently decrease abnormal joint loading of the intact limb.

Other physical therapy interventions to correct gait mechanics includes proprioceptive neuromuscular facilitation (PNF). Yiğiter et al examined the effectiveness of traditional prosthetic training and PNF resistive gait training in amputees in improving weight bearing and gait.[27] One group received traditional treatment including weight shifting, balance activities, stool stepping, and gait activities whereas the other group received the same activities with PNF.[27] Yiğiter et al report, "although the outcome of this study suggested that both therapeutic approaches were effective on weight bearing and gait biomechanics, better results were attained in the group who received proprioceptive feedback."[27] Both the Dingwell et al and Yiğiter et al studies indicate some form of feedback in gait training is beneficial in improving mechanics in amputees.

Strength Training

Addressing strength asymmetries will also improve joint loading during reciprocal gait. Amputees with preserved muscle strength demonstrate improved gait as a result of more control of the prosthesis and decreased energy expenditure.[28] Knee extensor, knee flexor, and hip abductor strength and symmetry correspond to improved function and symmetry in gait for amputees and should be targeted in rehabilitation.[13,29] Nadollek et al found a number of improvements were correlated with hip abductor strength including increased weight bearing on the prosthetic limb, improved gait parameters, and decreased medio-lateral center of pressure excursion of the prosthetic side.[14] Decreased weight bearing on the prosthetic limb is correlated with weak hip abductor strength, indicating the need for adequate strength in the gluteus medius and gluteus minimus.[14] In a study examining the relationship between tempro-spatial gait parameters and isometric strength, Boyd et al report, "results indicate that adequate force is necessary in both residual and sound limbs to improve functional gait ability."[29] A strength-training program targeting both strength and symmetry of strength between the residual limb and intact limb should be implemented early on to prevent asymmetrical gait leading to OA.

Other Treatments

Other preventative strategies include weight management, activity modification in vocational and recreational settings, and knee trauma prevention. A reduction in weight has been shown to significantly reduce the risk of OA by 50%.[5] Activity modification, such as altering activities requiring knee bending or carrying objects, reduces OA rates by 15-30%.[5] Pharmacology may also be indicated in some patients but should be done so with caution. Morgenroth et al state, "analgesic medications can increase the peak knee EAM [external adduction moment] secondary to pain reduction and can potentially accelerate OA progression in the long run." [8] That being said, a treatment that addresses mechanical factors should be fist considered in rehabilitation.

While total joint replacements in amputee patients remain rare they should not be ruled out in severe instances. When conservative treatments have been exhausted with no pain relief and continued deterioration in mobility, function, and independence, total joint replacements may be considered despite the added challenges that come with joint replacement rehabilitation in amputees. In a case report, one patient received a total knee replacement for grade 4 osteoarthritis in the intact limb following a below knee amputation 7 years earlier. [30] The patient received physical therapy post-operatively without limitations related to his amputation; at 6 weeks the patient presented improved function and was able to ambulate without assistive devices. [30] In another case report, a patient received a total hip replacement after having a contralateral hindquarter amputation and remained independent at follow up four years post-operatively. [31] Both cases indicate joint replacements can be preformed on amputee patients both safely and effectively when conservative treatments fail.

Conclusion

With the growing number of amputations and the increased prevalence of osteoarthritis in amputees, understanding the contributors and ways to prevent the development of joint degeneration in a population already facing mobility challenges is critical. Amputees ambulate with an asymmetrical gait creating an increase in joint loading on the intact limb resulting in bone remodeling and joint degeneration. Addressing temporal asymmetries through gait training with feedback, ensuring patients have proper prosthetic fit, alignment, and optimal prosthetic components, as well as providing strength training targeting hip abductors, knee flexors, and knee extensors all work to decrease the load in the intact limb. Each of these should be addressed early on for their preventative benefits to ensure optimal long-term mobility, participation, and quality of life.

Figures and Tables:
Temporal Asymmetry:[11]

Figure 1.

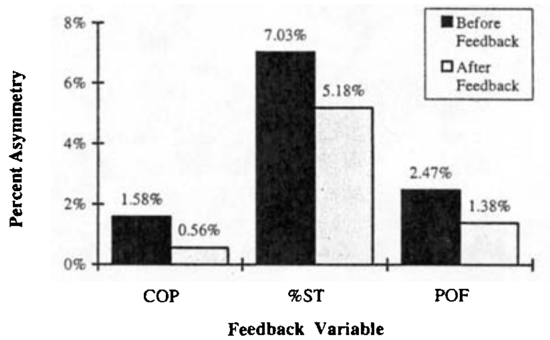

Fig. 3. Improvement in symmetry indices after real-time visual feedback training.

Figure 2.

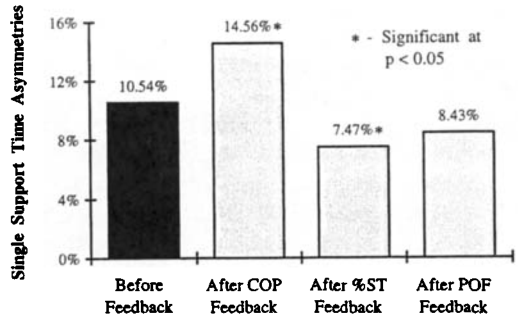

Fig. 4. Changes in single support time asymmetry (SISST) After real-time visual feedback training. (* – significantly different to before feedback at $p < 0.05$).

Muscle Strength Asymmetry:[13]

Figure 3.

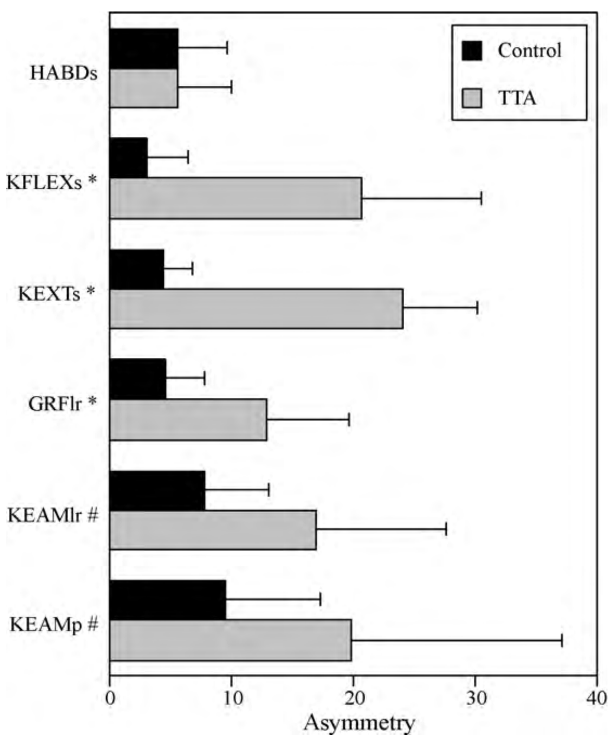

Fig. 2. Symmetry angle values and standard deviations in TTA (light gray) and control groups (dark gray). Larger values indicate greater asymmetry. See text for variable definitions. *Significantly more asymmetrical than the control group, $p < 0.05$; #effect size greater than 0.70.

Joint Loading: 17

Figure 4

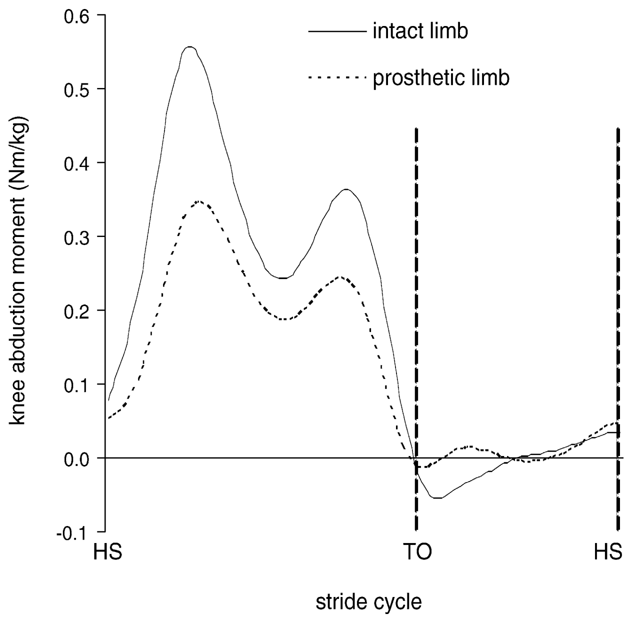

Fig. 1. The intact limb internal knee abduction moment was greater than the prosthetic side, indicating greater loading on the medial knee compartment. Data are the ensemble average of 10 subjects.

Figure 5

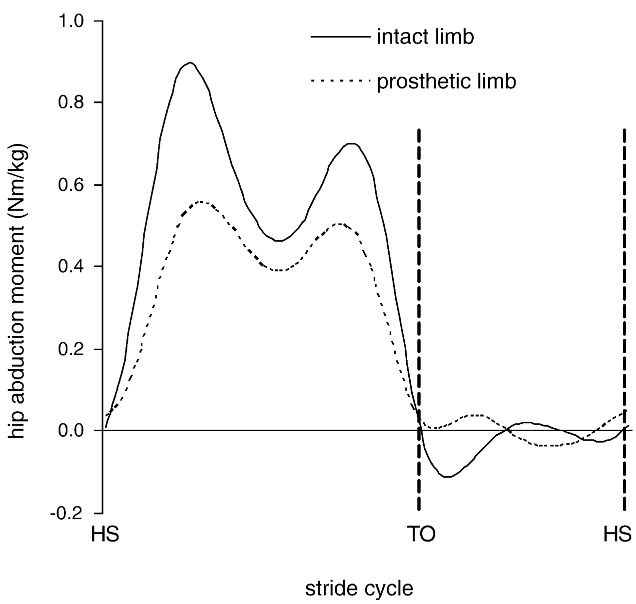

Fig. 2. The intact limb internal hip abduction moment was greater than the prosthetic side throughout stance. Data are the ensemble average of 10 subjects.

Amputation Level Joint Moments

Table 1[20]

Table 2. Peak net joint moments (Nm/kg) for the intact limb of trans-tibial, trans-femoral amputees and the left leg of able-bodied subjects walking at $1.2 m.s^{-1}$. PF=plantar flexor, DF= dorsiflexor, Ext=extensor, Flex=flexor moments.

Subject group	Net ankle moment		Net knee moment		Net hip moment	
	Ext(Nm/kg)	Flex(Nm/kg)	Ext(Nm/kg)	Flex(Nm/kg)	Ext(Nm/kg)	Flex(Nm/kg)
Able-bodied	1.17 (0.13)	0.05 (0.04)	0.79 (0.17)	0.23 (0.06)	0.71 (0.22)	1.04 (0.21)
Trans-tibial Intact limb	1.00 (0.04)	0.13 (0.07)	1.27* (0.28)	0.23 (0.64)	0.79 (0.23)	1.77* (0.41)
Trans-femoral Intact limb	1.29# (0.18)	0.16* (0.08)	1.14* (0.20)	0.29 (0.06)	1.14* (0.23)	1.22# (0.21)

* significantly differs from able-bodied (p<.05)
# significantly differs from trans-tibial amputees (p<.05)

Amputation Level Power Output

Table 2[20]

Table 3. Peak joint power outputs (W/kg) for the intact limb of trans-tibial, trans-femoral amputees and the left leg of able-bodied subjects walking at $1.2 m.s^{-1}$. Gen=power generated, Abs=power absorbed.

Subject group	Ankle power output		Knee power output		Hip power output	
	Gen (W/kg)	Abs (W/kg)	Gen (W/kg)	Abs (W/kg)	Gen (W/kg)	Abs (W/kg)
Able-bodied	1.22 (0.52)	1.21 (0.35)	0.47 (0.14)	2.87 (0.72)	0.85 (0.26)	0.85 (0.18)
Trans-tibial Intact limb	0.86 (0.41)	0.81 (0.47)	1.15* (0.28)	4.39 (1.59)	1.70* (0.40)	1.47 (0.53)
Trans-femoral Intact limb	1.74 (0.45)	0.99 (0.49)	1.04* (0.55)	4.40 (1.25)	1.01# (0.35)	1.29 (0.70)

* significantly differs from able-bodied (p<.05)
# significantly differs from trans-tibial amputees (p<.05)

Prosthetic Foot Component

Figure 6[21]

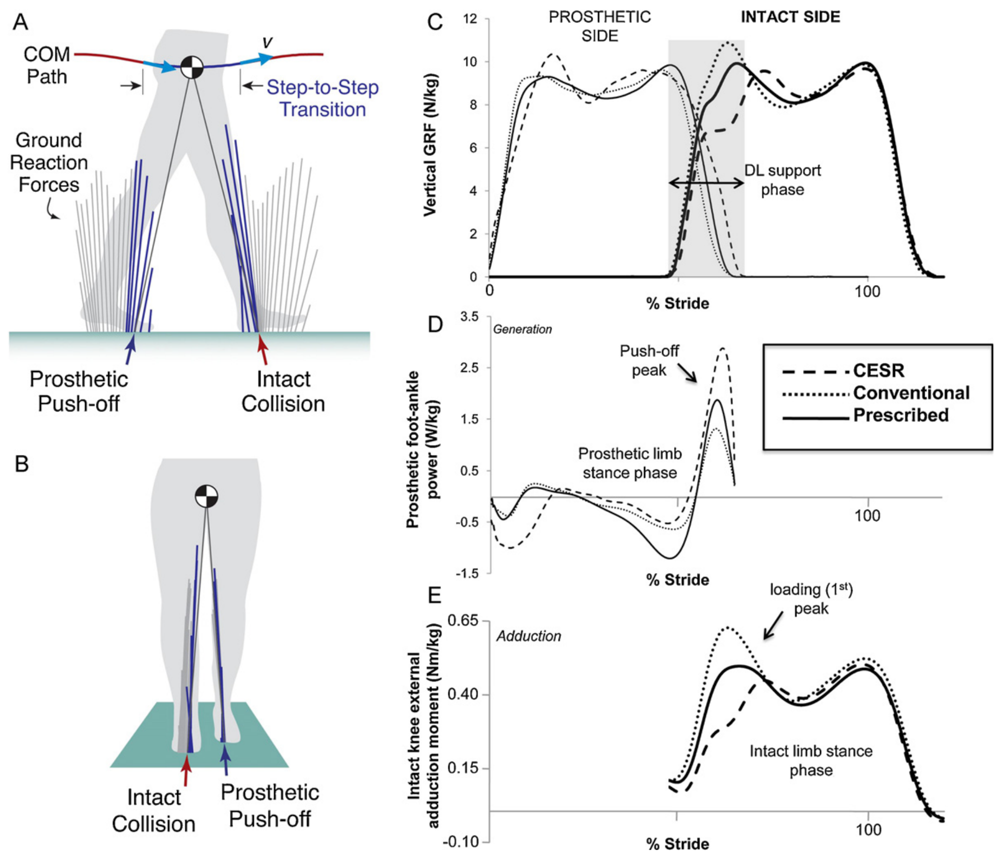

Fig. 1. Conceptual diagrams demonstrating the prosthetic and intact side ground reaction forces associated with the step-to-step transition (A and B) and graphic display of subject average vertical ground reaction force (C), prosthetic foot-ankle power (D), and intact knee external adduction moment (E), across the gait cycle for each prosthetic foot condition. (A) Sagittal view: the body center of mass (COM) is redirected during the step-to-step transition, as a result of forces applied by both legs against the ground (indicated by lines originating at center of pressure points throughout the gait cycle from a representative trial). (B) Coronal view: the ground reaction force is directed medial to the knee joint center of rotation essentially throughout stance phase leading to an external adduction moment. (C) Push-off and loading impulses were defined as the area under the vertical ground reaction force curves during double limb support. Prosthetic limb push-off impulse was negatively correlated with intact limb loading impulse. There was also a negative correlation between prosthetic foot-ankle push-off work (D) and intact limb 1st (loading) peak knee external adduction moment (E).

Prosthetic Foot Component

Figure 7[22]

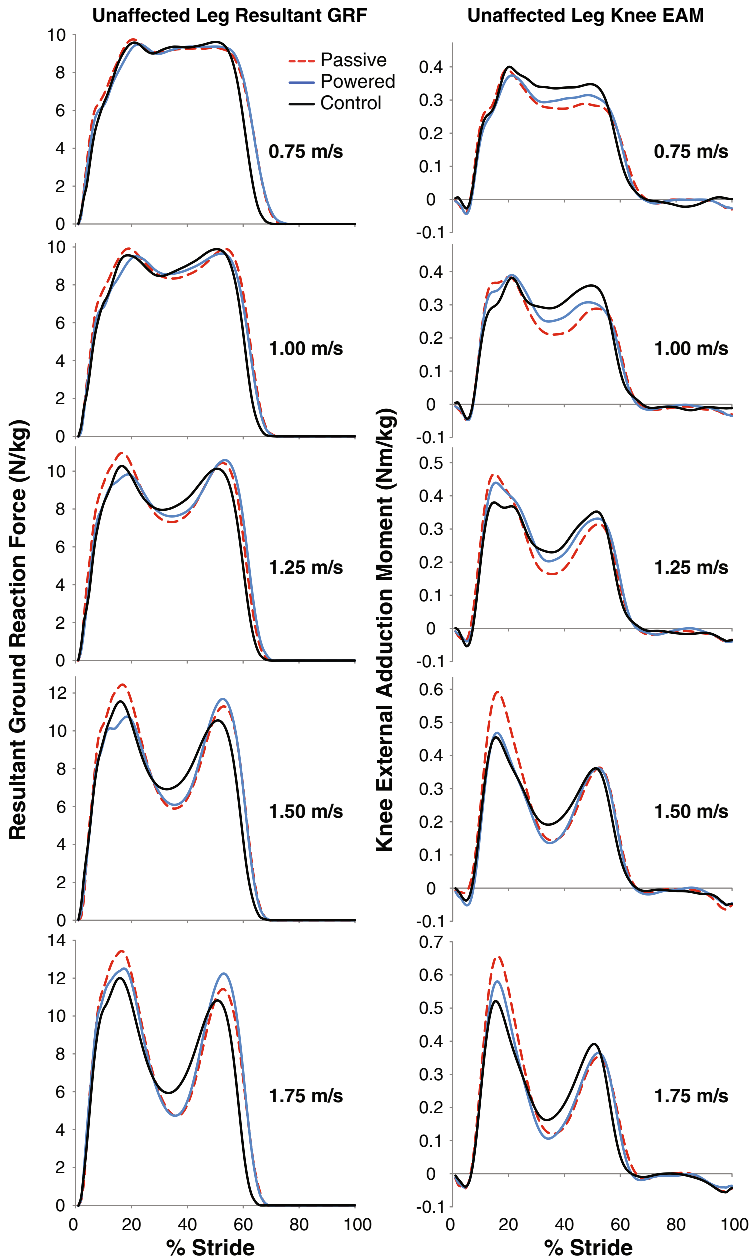

Figure 2 Average unaffected leg resultant ground reaction force (GRF) and knee external adduction moment (EAM). Dashed red lines indicate GRFs (left column) and EAMs (right column) of the unaffected leg while subjects walked using a passive-elastic prosthesis (Passive) across a range of speeds. Blue lines represent GRFs (left column) and EAMs (right column) of the unaffected leg while subjects walked using the powered prosthesis (Powered). Black lines represent GRFs (left column) and EAMs (right column) of non-amputees (Control). The average of three steps from all subjects is shown. Data are plotted versus percentage of a stride, where 0% occurs at heel strike.

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