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COMPENSATORY HIP AND KNEE MECHANICS IN TRANSTIBIAL AMPUTEES DURING STAIR DESCENT AND DIRECTIONAL TASK

Mindie Clark
minclark@nmu.edu

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By:

Mindie Clark

THESIS

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ABSTRACT

COMPENSATORY HIP AND KNEE MECHANICS IN TRANSTIBIAL AMPUTEES DURING STAIR DESCENT AND DIRECTIONAL TASK

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This study compared lower limb mechanics in unilateral transtibial amputees and able-bodied controls during strenuous activities of daily living (ADL). Seven unilateral transtibial amputee and five matched-abled bodied control participants executed stair descent on a four-step rehabilitation staircase followed by one of two anticipated directional tasks. Force, kinematics and gait parameters were chosen to compare mechanics and stride characteristics between the residual limb, intact limb and able-bodied dominant limb between a straight walking condition and a non-linear directional movement (wide-step cutting task). Results indicated that significant compensatory mechanisms occurred in the intact limb, perhaps from decreased load tolerance in the prosthetic limb. Compensatory mechanisms exhibited in the intact limb and the hip joint of the prosthetic limb exhibited mechanics that may indicate accelerated joint degeneration compared to able-bodied mechanics. Differences in the mechanics of limbs appear to be more pronounced during stair descent and a directional task compared level-ground walking. This may be a useful approach to identify and correct harmful mechanics.

KEY WORDS: Osteoarthritis, Force, Kinematics, Prosthetic

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This thesis follows format requirements specified by the School of Health and Human Performance at Northern Michigan University and the International Journal of Biomechanics and Movement Sciences, whose guidelines can be accessed at the link below.

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LIST OF SYMBOLS AND ABBREVIATIONS

Osteoarthritis…………………………………………………………………………...OA

Activities of Daily Living...……………………………………………………...…….ADL

Range of Motion……………………………………………………………………….ROM

Ground Reaction Force……………………………………………………………...GRF

Analysis of Variance...…………………………………………………………………ANOVA

LIST OF DEFINITIONS AS IT PERTAINS TO THIS THESIS

Able-bodied: A person without amputation

Trans-tibial: Below the knee, through the tibia

Contralateral: Opposite side or limb

Ipsilateral: Same side or limb

Intralimb/Intraindividual: Occurring within the same limb/individual

Interlimb/Interindividual: Occurring between separate limbs/individuals.

Joint Work: Measured in joules, the ability of components of a joint to produce power. Negative joint work is generally an absorption force, while positive work produces movement.

Joint moment: Measured in Nm, rotational torque exerted on joint by an intrinsic or extrinsic force

Range of motion: Measured in degrees, the movement potential of a particular joint

Eccentric: A muscle contraction that elongates as it produces force.

Concentric: A muscle contraction that shortens as it produces force.

CHAPTER I: JOURNAL MANUSCRIPT

INTRODUCTION

Osteoarthritis (OA) is the most common degenerative joint condition in the United States and is characterized by accelerated thinning of articular cartilage within a joint [25]. Prevalence of OA in lower-limb amputee populations is significantly higher compared to populations without amputation [22, 25]. In unilateral transtibial amputees, prevalence of OA in the hip of the amputated limb is 44%, compared to 15% in the intact limb and 11% in able-bodied populations [12]. Similarly, 41% of the transtibial amputee population report contralateral knee OA, compared to 22% among individuals with no amputation [12, 48]. Increased reliance on these joints during activities of daily living (ADL) has been suggested to influence the increased prevalence for OA in this population [23, 25]. Previous research has attempted to identify the relationship between limbs in unilateral amputees during gait and explain why increased reliance on specific joints occur [10, 12, 35, 36, 46]. Physical limitations associated with prosthetic design and reduced load tolerance of the residual limb appear to contribute to dissimilar mechanics between amputee and able-bodied populations [7, 46].

Improvements in prosthetic design has improved quality of life in amputee populations by reducing metabolic cost during ambulation through energy efficient prostheses [13, 29, 46]. Attempts to replicate the efficiency of a functional able-bodied ankle joint in prostheses has improved from 31% energy efficiency at 2.8 $m.s^{-1}$ in a conventional solid-ankle cushioned heel (SACH) foot to 84% in energy storing and returning devices [28, 47]. Energy efficiency has improved by incorporating an elastic keel to aid in propulsion during ambulation, which decreases dissipated energy through friction, termed hysteresis [14, 29].

Conventional socket design has influenced the knee joint of the residual limb to contribute minimally during ambulation. Decreased tolerance of loading in the residual limb knee joint during gait is speculated due to reports of pain, decreased force exerted during heel strike, and significantly longer stance time [10, 12, 30]. Decreased load tolerance of the residual knee joint prevents the execution of ADL and significantly increases mortality rate of amputee populations [12, 41].

Advancement in modern socket design has improved load tolerance of the residual limb by reducing pressure on bony landmarks susceptible to high pressure such as the distal end of the residual tibia and fibular head [15, 38]. Modeling a prosthetic socket after a pressure map of the residual limb allows for more evenly distributed loads throughout the limb and indicates less reliance on compensatory loads on the joints during ADL.

While load tolerance has improved with socket design, joint work of the residual limb knee joint seems reduced [7]. Sagittal plane knee range of motion (ROM) is restricted and ground reaction forces (GRF) are less than the knee joint of the intact limb and able-bodied mechanics during ambulation [7]. During difficult tasks such as stair ascent and descent, knee extensor strength is greatly reduced as well as reduced mobility of the residual knee joint [45]. To compensate for restricted knee joint work during various tasks, the ipsilateral hip joint increases ROM and muscle activity, even when performing tasks using the prosthetic limb as the lead limb [45]. Additionally, the intact limb demonstrates increases in work in all lower-limb joints, ROM, and gait parameters such as stride length [23, 32, 40]. Many of these compensatory mechanical differences are similar to populations who exhibit symptoms of predisposed OA at the hip and knee joints [25].

As the difficulty of movement increases, mechanical differences between amputee populations and their able- bodied counterparts become more pronounced [46]. Mechanical strategies used by amputee populations to perform strenuous ADL including stair descent and non-linear directional tasks increase the likelihood for injury and predisposition for OA [11]. During stair ascent, amputees exhibit faster walking speeds and greater limb asymmetry making them more susceptible to falling [45]. Additionally, stair tasks require more joint work exhibited by transtibial amputees [8]. Because the ankle joint produces limited power even with an energy-storing prosthetic, most work comes from the hip joint. [32]. Performing non-linear directional tasks such as turning a corner require strong hip and knee extensors to stabilize the body due to compromised knee stability in the frontal plane [16, 22].

While past research has compared mechanical differences during walking and stair ascent and descent in amputee populations, other strenuous ADL have yet to be compared. Stair descent and non-linear directional tasks may elicit strategic multi-planar movement in joints of the lower limb in able-bodied populations [16, 32]. The purpose of the current study was to compare differences in mechanical strategies used by individuals with transtibial amputation and matched, able-bodied counterparts during of stair descent and a non-linear directional task. It is unknown how the knee joint of the amputated limb will contribute during these strenuous tasks, and how the ipsilateral hip will compensate for the speculated reduction in knee joint work. If the ipsilateral hip joint compensates more during these ADL, joint degeneration may be accelerated, increasing the need to develop a more functional prosthetic socket that mimics able-bodied mechanics; thus, improving quality of life for amputee populations.

METHODS

Participants

Seven males with unilateral transtibial amputations volunteered to participate in the current research study. Participants were included if they were able to effectively perform the protocol, ambulate without an assistive walking aid, and use a well-fitted prosthetic device with a perceived socket slip of less than 6mm [24]. Five age-matched able-bodied participants were recruited to compare mechanics of only their dominant limb to both limbs of amputee participants.

Following the approval from the Institutional Review Board to complete the current study, anthropometrics were measured following the completion of the informed consent (Appendix A) as well as a participant survey regarding ADL functionality, pain, numbness, amputation and leg dominance (Appendix B) [33]. Independent t-tests indicated no significant differences between groups, displayed in Table 1.

Protocol

Participants were familiarized with the protocol in which a self-selected pace was controlled throughout data collection using a metronome matched to the desired cadence. Participants began each trial on the top platform of a four-step wooden rehabilitation staircase with handrails (Bolingbrook, IL, USA) (Figure 1). Participants were permitted to use handrails as needed while they descended the staircase onto one AMTI force platform (OR6-2000 Advanced Mechanical Technology, INC. [AMTI], Watertown, MA, USA) and continued in a fluid motion for two gait cycles in one of two anticipated directions. Participants were instructed prior to stair descent to continue straight ahead for two full gait cycles or side step 45 degrees to the contralateral side [39]. Colored tape was placed at 35 and 55 degrees away from the center of the force platform to aid in directional targeting (Figure 2). Familiarization trials were performed in each direction beginning with either foot as needed until the participant verbalized confidence completing each direction with the desired cadence.

Directional tasks were randomized block for each amputee participant and the order was replicated by matched- control participants. Participants completed 12 successful cadence-controlled trials. Three trials in each direction, beginning with each foot, were determined successful if the participant's entire foot contacted the force platform and continued in the desired direction for two gait cycles.

Markerset

6- mm reflective markers were placed on participants' bony landmarks and surfaces to measure kinematics of the lower limbs (Figure 3). Markers were placed bilaterally on participants' anterior superior iliac spines, posterior superior iliac spines, sacrum, greater trochanters, medial and lateral knee joint lines, lateral malleoli, medial malleoli, head of the fifth metatarsal, and superior aspect of the big toe. Additionally, marker clusters were placed on the lateral thigh and shank halfway between markered bony landmarks to measure axial rotation [13]. Markers on the amputated limb were estimated from bony landmarks of the participant's intact limb (Figure 3). Markers for the ipsilateral knee joint line were placed on the prosthetic socket at the point of sagittal motion.

Variables

Ten Cortex Motion Analysis Corporation cameras (Santa Monica, CA, USA) were placed around the capture space to collect 3-dimensional data during trials at 250 Hz. Biomechanical variables were chosen to compare mechanics of participant's prosthetic limb, intact limb, and the matched- control participant's dominant limb throughout the stance phase of the limb on the force platform from touchdown to take-off. Hip and knee ROM in all planes and gait parameters including step length and stance time were calculated using Visual 3D software (Germantown, MD, USA). Data were tracked and filtered using fourth order Butterworth filter with a cutoff frequency of 6 Hz with Cortex Motion Analysis (Santa Monica, CA, USA). Force data were collected at 750 Hz to measure peak vertical ground reaction force (GRF_{peak}) and normalized to body weight (N/Kg). Moments of hip and knee joints in all planes were calculated using Visual 3D software (Germantown, MD, USA).

Statistical Analyses

Participant characteristics were compared for normal distribution using descriptive statistics in SPSS v.24 software (SPSS Inc, Chicago, IL). Force, gait parameter, and mechanical variables were averaged across trials of each condition- straight walking and non-linear directional task, and compared between three limbs- prosthetic, intact, and matched able-bodied dominant limbs- during the stance phase on the force platform. This comparison was calculated using a two-way mixed ANOVA with a 2-by-3 comparison between the two directional tasks and three limbs. The leg factor had three levels in which all three combinations of legs were compared. The directional tasks had two levels between the straight-walking task and non-linear wide step task. Pairwise comparisons were used to identify differences between specific legs during specific conditions, using a Bonferroni adjustment for multiple comparisons. Partial eta squared (np^2) effect sizes were used to determine magnitude of differences using the following classification: <0.04 = trivial, 0.041 to 0.249 = small, 0.25 to 0.549 = medium, 0.55 to 0.799 = large, and >0.8 = very large [18]. Alpha was set at $p<0.05$.

RESULTS

Forces

No significant differences and small effect sizes were seen between groups or conditions for GRF_{peak} or normalized forces (p>0.05). Although the prosthetic limb exhibited lower GRF_{peak} compared to the intact and able-bodied limbs, differences were not significant (Table 2). Timing of GRF_{peak} were not significantly different between groups or tasks (p>0.05). There were no significant interactions between group and tasks for force data.

Gait Parameters

Significantly less stance time was seen only during the non-linear task between the able-bodied limb compared and the intact limb (p=0.033, np^2=0.261), but not the prosthetic limb (p>0.05), shown in Table 3.

Stride length was not significantly different between limbs or tasks (p>0.05). There were no significant interactions between groups and tasks for gait parameters.

Hip Mechanics

No significant differences and very small effect sizes were found regarding sagittal plane hip mechanics between limbs or tasks (p>0.05), shown in Figure 4.

Hip joint adduction angles (P=0.045, np^2=0.380) were significantly greater with a medium effect size between limbs in the wide-step non-linear task. The able-bodied limb indicated significantly greater hip joint adduction compared to the prosthetic limb (p=0.022), but not the intact limb (p>0.05). No significant differences were found in hip abduction angles or frontal plane hip ROM (Figure 5).

Hip joint adduction moments indicated significant differences and a medium effect size between limbs and groups (p=0.018, np^2=0.460). Pairwise comparisons showed that hip adduction moments were greater in the able-bodied limb compared to both prosthetic ($p=0.006$) and intact limbs ($p=0.028$) during the wide-step non-linear task.

Transverse plane hip movement did not indicate significant differences between groups or tasks ($p>0.05$) (Figure 6). All limbs exhibited significantly greater hip external rotation moments during the wide-step non-linear tasks ($p=0.046$, $np^2=0.272$). Hip internal rotation moments were significantly greater during the non-linear directional task in the able-bodied limb compared to the prosthetic limb ($p=0.044$). No significant differences were seen during the straight walking task. There were no significant interactions between groups or tasks for hip or knee mechanics.

Knee Mechanics

Significantly greater knee ROM in the sagittal plane existed during the straight walking task compared to the non-linear task in all limbs, with a medium effect size ($p=0.028$, $np^2=0.319$). No significant differences and a large interindividual variance was indicated for peak extension or flexion angles, although a medium effect size was seen in peak knee flexion between limbs ($p=0.055$, $np^2=0.255$) (Figure 7).

No significant differences and a large variance was found between limbs and tasks for frontal and transverse knee joint mechanics ($p>0.05$) (Figure 8).

DISCUSSION/CONCLUSION

The purpose of the current study was to compare the mechanics in the lower limbs of transtibial amputees during strenuous ADL such as stair descent and a directional cutting task. The knee joint in a transtibial-amputated limb has been shown to contribute minimally to walking gait [7]. Strenuous ADL such as stair descent and non-linear directional tasks may highlight the mechanical strategies used by this population to compensate for the limited contribution of the amputated-limb knee joint [33, 43, 45]. These potential compensations may provide an explanation for the increased prevalence of OA among populations with transtibial amputation [35, 43].

Force

Past research has indicated a marked asymmetry in GRF_{peak} between both limbs in unilateral transtibial amputees, which may partially explain the incidence of OA in the intact-knee [10, 11, 25]. Greater force differences between the amputated limb, intact, and able-bodied limbs were reported in past research of 12-20% during level-ground ambulation, compared to force differences seen in the current study [46]. Furthermore, a 54% difference in forces between limbs of an amputated individual have indicated predisposition of OA, which was substantially greater than the force differences seen in the current study [25]. A greater interindividual variance was seen in the current study among all limbs, which may explain the lack of significance.

The increased difficulty of weight acceptance onto a force platform following stair descent typically elicits greater GRF_{peak} in amputee populations compared to level-ground ADL, which was not seen in the current study [32, 40, 45]. However, the amputee group appeared anecdotally to implement cautious strategies to lessen such forces during stair descent including the use of handrails and attempts to control the loading rate by using eccentric muscle activity on the contralateral limb. Although these variables were not quantified in the current study, both have been strategies used by this population reported in previous research [32, 45]. Such strategies used to execute strenuous ADL may be used by amputees regardless of the starting limb, which may decrease force asymmetry. Force asymmetry is not indicated in functional able-bodied populations but may be a consideration to include as a comparison to amputee limb asymmetry in future studies.

Gait Parameters

Reductions in stride length have been reported in amputee populations compared to able-bodied populations, and asymmetrical stride length has been reported in amputee gait [10, 11, 23]. Stride length was not significantly different between limbs or directional tasks in the current study, which may be due to fixed directional target angles. The distance from the staircase to the force platform was adjusted for each participant to ensure his entire foot contacted the platform, which may reduce variability in stride length.

Stance time was 16 and 17% longer in the intact limb compared to the prosthetic and able-bodied limbs, respectively, although stance time was only significantly different between the intact and able-bodied limbs. This is a greater difference than those reported in level-ground ambulation [23] but similar to previous reports measuring gait parameters in amputees during stair tasks [33]. These comparisons may suggest that greater asymmetries occur during more demanding tasks, similar to the observation that greater mechanical dissimilarities occur between populations at greater speeds [46]. Greater stance time on the intact limb indicates a possible compensatory mechanism due to reduced load tolerance of the prosthetic limb [38]. Increased reports of pain and pressure on bony landmarks may cause amputees to implement strategies to spend less time on the residual limb [11, 15]. Lack of proprioception while the prosthetic limb is in swing phase may require more time to complete a directional task compared to a sound limb, explaining the increased stance time on the intact limb [45]. The ability to efficiently execute a strenuous ADL may improve over time in amputee populations, which may explain the high variability currently seen in gait parameters among the amputee group. Increased stance time on the intact limb may lead to an asymmetrical increase in loading, which may explain accelerated joint degeneration in the amputee’s intact limb [23].

Mechanics

Hip adduction movements and adductor moments were significantly greater in able-bodied populations, which has previously indicated to reduce risk for OA of the lower limb. While hip abduction moments were not greater in amputee participants, higher OA prevalence in amputee populations may be explained by a lack of hip adduction during strenuous tasks [19]. Efforts to move the limb closer to the midline in amputee participants may not have been practical when executing the strenuous ADL because of a wider base of support needed to stabilize participants, which elicits greater hip abduction [25].

The primary stabilizing mechanisms that the able-bodied participants appeared to employ in the current study were the increased moments at the hip, which may help offset trunk deviation. This was not evident in the amputee group, who may have chosen inefficient strategies to stabilize such as the use of handrails to lower the limb to the force platform. Decreased intrinsic stability of the prosthetic limb along with previously reported reduction in load tolerance of the prosthetic limb knee joint may contribute to compensatory loading asymmetry of the intact limb. [11, 46].

Forward trunk flexion is typically displayed during amputee gait to allow for visual strategies of foot placement during walking tasks. This is suggested to be caused by the lack of proprioception in the amputated limb [11, 45]. Trunk flexion may reduce sagittal hip ROM, although this variable was not measured in the current study. The protocol of the current study required accurate foot placement, which may have induced greater trunk flexion to visually strategize foot placement. If this was the case, improved neuromuscular components of the trunk and hip may be a consideration to improve stability of the amputated limb [19].

Knee mechanics were not significantly different between limbs in the current study, primarily due to a large variance in all planes of motion. Past reports of mechanical differences of the knee joint typically assess a full gait cycle, whereas the current study only measured mechanics during the stance phase. While more knee sagittal plane ROM occurs during the swing phase of a limb, perhaps mechanical dissimilarities between amputee and able-bodied populations occur primarily in the swing phase of gait.

Directional Tasks

The anticipatory aspect of the current protocol may have influenced hip mechanics. Increased external hip rotation moments may have occurred to provide greater stability during the wide-step task when the swing limb executed a 45- degree step to the opposite side. Additionally, less sagittal hip ROM was seen during the non-linear wide-step task which may have indicated that hip flexors and extensors are used as stabilizers during non-linear tasks, when frontal and transverse work are needed to a greater extent to execute movement.

The current protocol was designed to reflect more strenuous ADL but directional tasks were purposefully anticipated to prevent injury [45]. However, comparisons to unanticipated non-linear movement may be a future research consideration to highlight problem-solving mechanical strategies in amputee populations.

Conclusion and Limitations

Executing strenuous ADL such as stair descent and non-linear directional tasks elicit mechanisms in unilateral transtibial amputees that may indicate a heightened predisposition for OA [11, 23, 40]. Heightened differences in mechanics used by amputee populations compared to able-bodied populations appear to indicate harmful effects that accelerate joint degeneration [25, 32]. Training proper gait mechanics in amputees may reduce asymmetries and provide stability when executing strenuous ADL.

The current study limited the comparison of mechanics to the lower limbs. Trunk position, the use of handrails, and muscle activity of the contralateral limb on the lowest step should be quantified in future research, as these variables were speculated to be more prevalent in the execution of tasks in amputee participants.

Table 1. Characteristics of amputee and able-bodied control participants displayed as mean ± SD.

	Amputee (n=7)	Control (n=5)
Age (yr)	54.14 ± 17.35	55.40 ± 22.09
Height (cm)	187.90 ± 7.16	180.79 ± 6.42
Weight (kg)	102.57 ± 26.33	92.19 ± 18.40
Duration of amputation (yr)	11.88 ± 19.01	

Table 2: Peak GRF (N) ± SD, GRF normalized (N/Kg) ± SD for legs and conditions.

Task	Leg	Force (N) ± SD	Normalized Force (N/Kg) ± SD
Straight	Amputated	1257.56 ± 314	12.62 ± 2.78
	Intact	1502.70 ± 425	15.00 ± 3.22
	Able- bodied	1446.73 ± 453	16.07 ± 6.13
Directional	Amputated	1217.88 ± 255	12.19 ± 2.05
	Intact	1589.42 ± 405	15.99 ± 3.91
	Able- bodied	1496.45 ± 494	16.48 ± 5.74

Table 3: Mean ± SD for stance time (seconds) between limbs and tasks.

Task	Leg	Stride Length (m)	Stance Time (sec)
Straight	Amputated	0.67 ± .07	0.65 ± .22
	Intact	0.61 ± .14	0.85 ± .03
	Able- bodied	0.71 ± .05	0.67 ± .19
Directional	Amputated	0.65 ± .11	0.79 ± .12
	Intact	0.72 ± .17	0.92 ± .21
	Able- bodied	0.73 ± .06	0.71 ± 03*

* Indicates significant differences when compared to the intact limb.

Table 4: Mean ± SD hip moments in all planes, normalized to participant's body weight (Nm/kg).

Task	Leg	Flexion	Extension	Abduction	Adduction	Internal Rotation	External Rotation
Straight	Amputated	0.425 ± .29	-0.414 ± .27	0.557 ± .82	-.471 ± .59	-0.152 ± .36	0.267 ± .40
	Intact	0.498 ± .15	-0.365 ± .29	0.367 ± .35	-.727 ± .87	-0.421 ± .49	0.022 ± .18
	Able- bodied	0.657 ± .11	-0.552 ± .47	0.128 ± .07	-1.607 ± .63	-0.548 ± .47	0.039 ± .08
Directional	Amputated	0.289 ± .20	-0.428 ± .23	0.435 ± .75	-0.417 ± .49	-0.118 ± .36	0.325 ± .45$^{\#}$
	Intact	0.393 ± .22	-0.428 ± .18	0.288 ± .27	-0.717 ± .63	-0.343 ± .30	0.051 ± .22$^{\#}$
	Able- bodied	0.554 ± .14	-0.621 ± .44	0.133 ± .09	-1.662 ± .68^{*o}	-0.670 ± .52	0.089 ± .07$^{\#}$

* Indicates significant differences when compared to the intact limb.

^{o} Indicates significant differences when compared to the amputated limb.

$^{\#}$ Indicates significant differences when compared to the straight task.

Table 5: Mean ± SD knee moments in the sagittal and frontal plane, normalized to participant's body weight (Nm/kg).

Task	Leg	Flexion	Extension	Abduction	Adduction
Straight	Amputated	-0.179 ± .16	0.367 ± .27	0.201 ± .27	-0.202 ± .27
	Intact	-0.069 ± .30	0.685 ± .29	0.146 ± .14	-0.286 ± .39
	Able- bodied	-0.079 ± .32	0.728 ± .45	0.174 ± .23	-0.397 ± .39
Directional	Amputated	-0.163 ± .14	0.362 ± .23	0.133 ± .23	-0.179 ± .24
	Intact	0.095 ± .57	0.789 ± .18	0.146 ± .08	-0.239 ± .42
	Able- bodied	-0.149 ± .38	0.768 ± .41	0.050 ± .03	-0.415 ± .48

Table 6: Mechanics associated with amputee gait according to previous research [11, 33, 40, 45].

		Kinematics	Kinetics	Gait Parameters
Knee	Residual	< Knee Flexion*	> Flexor and Extensor Negative work*	> Stance time*
		>Time to peak flexion	< Flexor/Extensor Positive work	< Step length*
		< Frontal ROM	> Duration of flexor/extensor work	< Swing time
			< VGRF	
	Intact		> VGRF*	> Step length
			> Extensor work	
Hip	Residual	> Flexion	> Positive and negative work, flexor/extensors*	> Trunk Flexion*
		< Extension*	> Negative work, frontal plane*	
		> Frontal ROM*		
		> External Rotation		
	Intact	> Adduction		

*Associated with OA predisposition [19, 23, 25, 43]

Figure 1: Rehabilitation staircase with handrails.

Dimensions: total width: 76.2 cm, total length: 139.7 cm, total height: 137.2 cm.
Dimensions for lowest step: width: 76.2 cm, length: 25.4 cm, height: 25.24 cm.

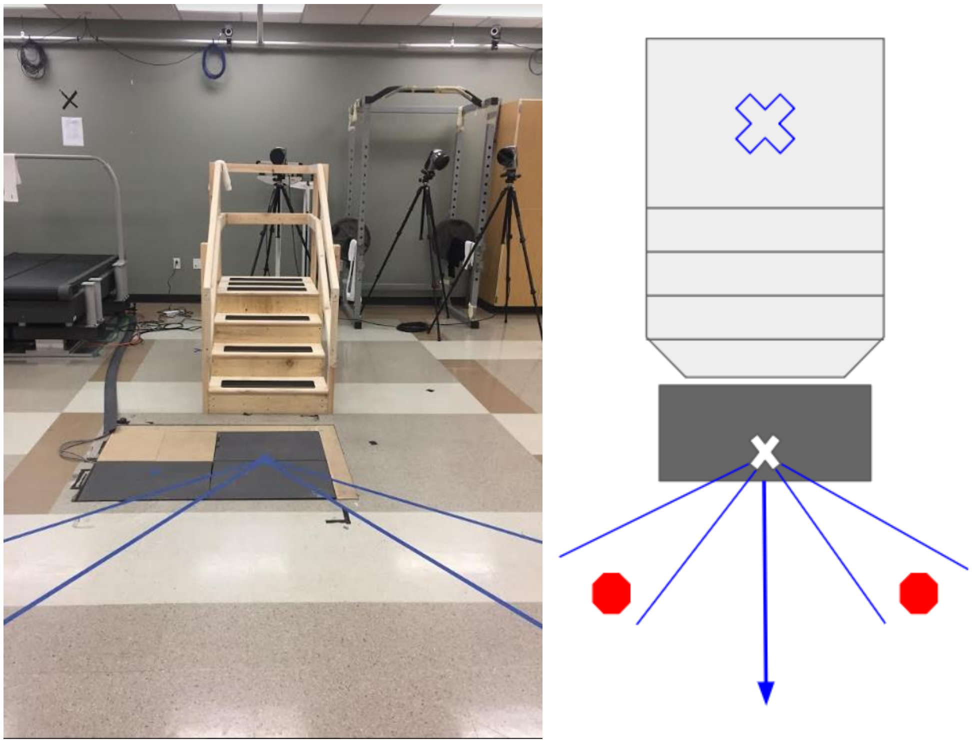

Figure 2: Protocol set-up, frontal and overhead view: Staircase, force platform and colored-tape trail.

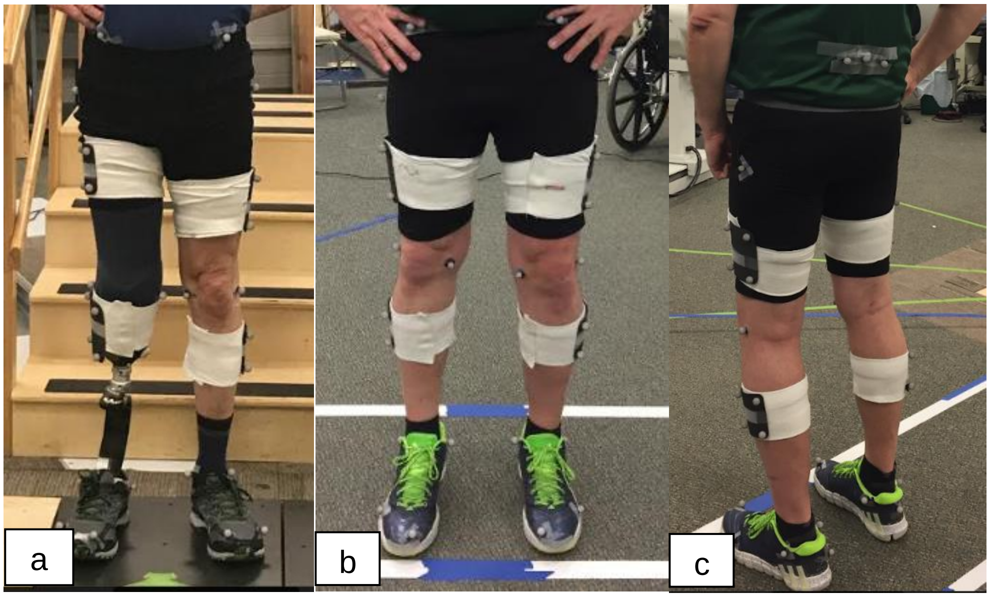

Figure 3: Marker placement on the lower limbs for:

a. Unilateral amputee, anterior view.
b. Matched control, anterior view.
c. Matched control, posterolateral view.

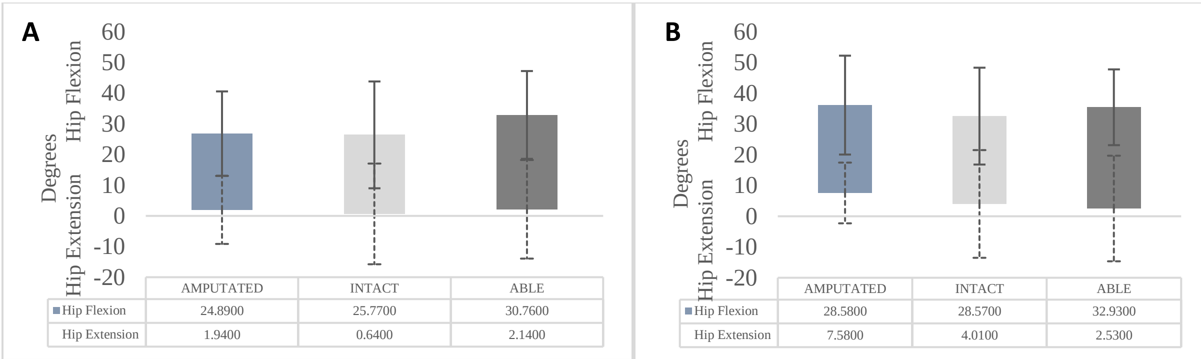

Figure 4: Sagittal plane hip movement between legs and tasks (A: straight, B: directional). The colored box represents ROM for the respective leg during the tasks, with peak flexion and extension indicating the location of the box. SD are represented for each peak value: ---- represents SD for peak hip extension, —— represents SD for peak hip flexion.

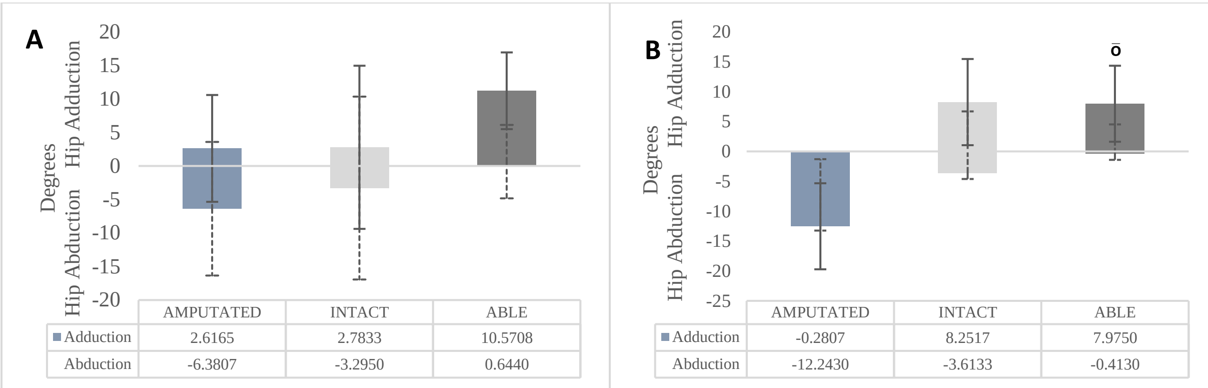

Figure 5: Frontal plane hip movement between legs and tasks (A: straight, B: directional). The colored box represents ROM for the respective leg during the task, with peak adduction and abduction indicating the location of the box. SD are represented for each peak value: ---- represents SD for peak hip abduction, —— represents SD for peak hip adduction.

º Indicates significant differences when compared to the amputated limb.

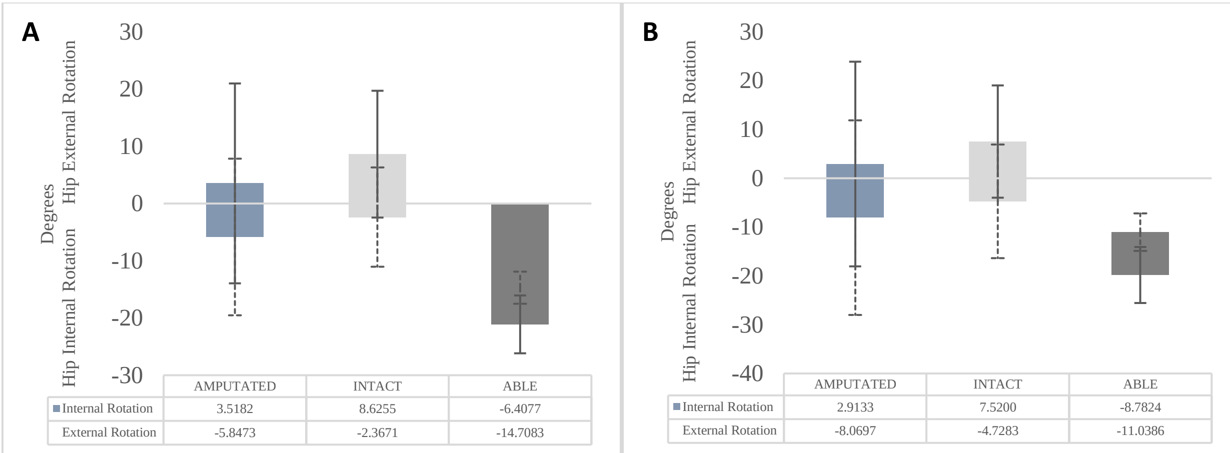

Figure 6: Transverse plane hip movement between legs and tasks (A: straight, B: directional). The colored box represents ROM for the respective leg during the task, with peak external and internal rotation indicating the location of the box. SD are represented for each peak value: ---- represents SD for peak hip internal rotation, —— represents SD for peak hip external rotation.

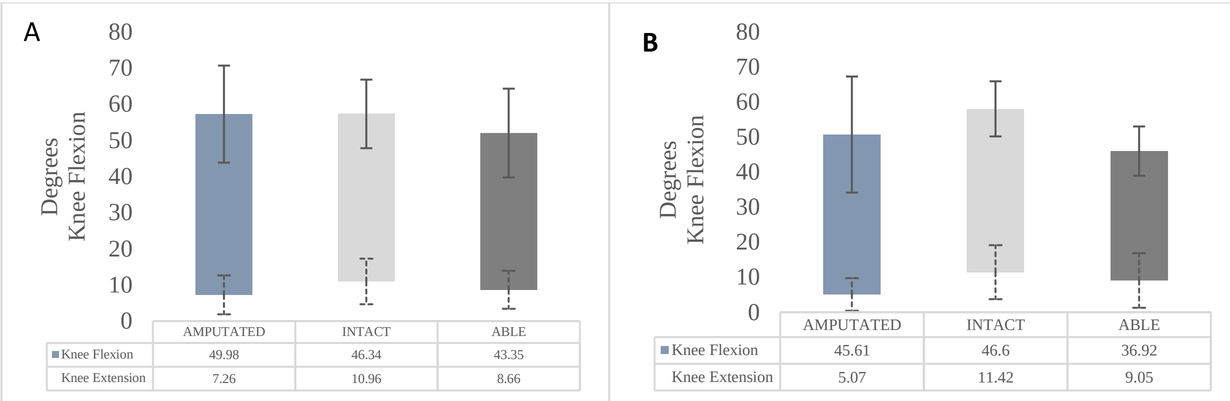

Figure 7: Sagittal plane knee movement between legs and tasks (A: straight, B: directional).The colored box represents ROM for the respective leg during the task, with peak flexion and extension indicating the location of the box. SD are represented for each peak value: ---- represents SD for peak knee extension, — represents SD for peak knee flexion.

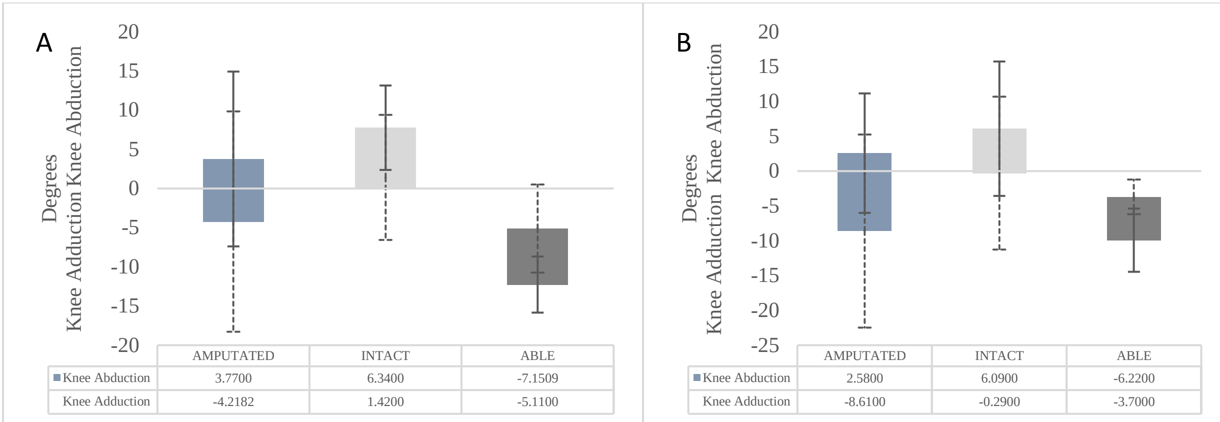

Figure 8: Frontal plane knee movement between legs and tasks (A: straight, B: directional). The colored box represents ROM for the respective leg during the task, with peak adduction and abduction indicating the location of the box. SD are represented for each peak value: ---- represents SD for peak knee adduction, —— represents SD for peak knee abduction.

CHAPTER II: LITERATURE REVIEW

Planes of Movement

Human movement occurs in three planes: sagittal, frontal and transverse. These planes are required for properly executing activities of daily living (ADL). When one or more of these planes are exaggerated or restricted, abnormality of biomechanics indicate an increased risk for injury [11, 40]. Osteoarthritis (OA) is a secondary condition that commonly occurs at a higher rate as a result of mechanical abnormalities in populations such as previously injured and amputated individuals [6, 25, 30]. Efforts to limit and correct abnormal mechanics used to carry out ADL in these populations present an ongoing need for research to prevent subsequent injury and improve quality of life [23, 27].

ADL in Research

Walking is an ADL commonly used in research to compare differences in mechanics between populations of various function in all planes of motion because it is involved in many ADL and can be compared easily [44, 47]. Energy cost and differences in kinematics and kinetics can be compared during walking between clinical, athletic and able-bodied populations [13, 25].

Decreased stability has been previously explained a negative correlation between walking speed and level of functionality [20]. Risk of falling has also been positively correlated with speed of executing ADL such as walking and stair tasks [45]. Individual joint kinematics and kinetics may serve as more specific variables to represent how clinical populations differ while performing ADL [32].

Among highly functional amputee athlete, greater mechanical differences were seen with greater running speeds in track events [46] despite the use of more energy efficient prosthetic devices [28]. Metabolic cost was higher for amputee athletes, even when accounting for a lighter limb, suggesting inefficient mechanics, possibly to compensate for a less responsive prosthetic limb [8, 10, 46].

If metabolic cost is increased in amputee populations to compensate for structural limitations of a prosthetic device, mechanical disadvantages may indicate negative long-term effects due to compensatory mechanisms elsewhere in the body [27, 30, 40]. Furthermore, if higher speeds elicit greater dissimilarities in mechanical strategies to execute a task, perhaps mechanics used to perform ADL of greater difficulty are more pronounced [21, 32].

Ankle Joint

Replicating an ankle joint is one of the most difficult tasks in prosthetic design. The bones and musculature of the ankle joint aid in ambulation more so than any other joint in the body [29]. The role of the sound ankle joint is to control movement, absorb, and generate energy during ambulation [8]. The heel and plantar-flexors absorb energy as the heel strikes the ground. Subsequently, dorsi-flexors eccentrically contract to control the movement of one's toe to the ground.. Plantar and dorsi-flexors control the movement of the tibia over the foot during midstance, with a large generation of energy during the contraction of the plantar flexors to initiate toe-off [9, 44, 47].

Energy efficiency of 100% means that the same amount of energy that is stored is returned. In materials that lack generative components, an inherent amount of stored energy will be dissipated by friction or interference during movement [28]. Because an able-bodied ankle joint has a substantial ability to generate energy due to ankle dorsi and plantar-flexor muscles, hysteresis is negated [46].

Because prosthetic ankle joints rarely have a generative component, energy efficiency is difficult to replicate [8, 14, 29]. Energy storage of an able-bodied ankle joint is calculated by multiplying ankle joint angular velocity and net ankle joint moments [28]. The axis of rotation at a prosthetic ankle joint is unclear and presents inconsistencies when calculating energy efficiency using the traditional method [28].

A prosthetic heel, much like the heel of an able-bodied foot, is designed to compress and absorb force during heel strike and provide stability through gait [14]. A prosthetic heel has not changed significantly since the 1950's when the Solid Ankle Cushioned Heel (SACH) foot was designed, and is often made of a stiff compressive foam [14].

The minimized elasticity of the SACH heel contributes little to the efforts made to replicate the energy-generating component of a sound ankle joint [9]. Originally, the prosthetic keel, which represents the individual's ankle joint, was made from rigid material but has since progressed to elastic spring-like materials to elicit a dynamic response during gait [9, 29]. The modern-day keel is designed to compress when loaded during heel strike and midstance. This energy is subsequently released to assist in propulsion of the limb during swing phase [32].

Advancement in technology of the prosthetic keel has significantly improved the quality of life of amputee populations by improving efficiency from approximately 30% in the solid-ankle-cushioned-heel (SACH) foot at 2.2 m.s-1 to 60% in a dynamic-response energy- storing foot and 80% in sprinting prosthetics at 9.25 m.s-1 [4, 14]. A large discrepancy remains regarding the energy efficiency of a sound ankle joint compared to prosthetics because of the prosthetic's inability to generate energy to the extent of sound ankle joint musculature [28]. Additionally, high-energy efficiency is only elicited when a high load is placed on the device seen during running and sprinting, but has not been replicated at high efficiencies in devices worn for daily living [4].

Knee Joint

Significant mechanical differences are seen between the knee joint of an amputated limb and an able-bodied knee joint. Transtibial amputees lack a sound ankle-joint, which results in a transfer of load during gait to the residual limb [36]. The soft tissues on the residual limb are less tolerant to forces associated with heel strike during ambulation [30, 38]. To compensate for this reduced load tolerance, contact time is longer on the amputated limb to allow for weight acceptance, and ground reaction forces (GRF) are reduced 22% on the amputated limb [10, 46]. Additionally, the intact limb of unilateral amputees exhibits reduced frontal plane movement, greater GRF and increased stride length as efforts to compensate for the decreased load tolerance of the residual limb [40, 43].

Technological advancements in prosthetic socket design and material have improved quality of life of amputees by delaying residual limb joint degeneration. During gait, shear forces occur on high pressure areas of the residual limb, namely the fibular head and the distal end of the tibia [15]. When socket material is more compliant based on the pressure measurements in the limb, pressure decreases during gait by 15-17% and 7-8% on the fibular head and distal tibia, respectively, despite greater self-selected walking speeds [38]. Additionally, pressure of the fibular head decreases by 13-21% during load-bearing static movements using a compliant socket, despite a 3-fold weight increase in socket design [38]. This technological advancement indicates that loading during weight-bearing activities is distributed more evenly throughout an amputated limb. This enables and motivates the individual to execute ADL more efficiently [12, 41].

Although improvements in socket material has allowed for tolerance of weight-bearing activities, prosthetic sockets restrict mobility of the residual knee joint [7, 32, 33]. The residual limb is inserted into the prosthetic socket to provide an extension for the amputated limb via the attached device; while the patellar tendon-bearing socket comes above the femoral condyles. Because the socket extends above the joint line, mobility does not reach full potential and reduces knee flexion during stance [33]. Reduced sagittal plane range of motion (ROM) during gait reduces sagittal knee moments during walking [7, 8].

Proper fitting of the components of a prosthetic device determine the ability of an amputee to ambulate effectively [27]. Proper fitting of the prosthetic socket is difficult to maintain over time because the residual limb changes in size due to swelling and tissue makeup [12]. Efforts to reduce socket slip to less than 6mm have improved knee mechanics of the amputated limb and should be regulated over time [24, 27].

In addition to proper socket fit, alignment is a modifiable determinant of one's ability to ambulate with a prosthetic device. Prosthetic alignment refers to the relative position of the prosthetic foot and pylon relative to the rest of the body [12]. As an amputee begins to ambulate, prosthetic alignment is more lateral to provide greater stability but puts excess stress on the medial component of the residual knee joint [27]. Malalignment in the frontal and sagittal planes within 10 degrees does not appear to significantly alter gait for long-term prosthetic users [31]. However, mechanical integrity of the knee joint becomes compromised beyond 15 degrees of malalignment, most significantly in the frontal plane [3]. This may negatively affect frontal hip and knee moments, which are associated with OA

While improvements in prosthetics have increased quality of the residual limb knee joint, contribution of this joint during ambulation is minimal [7, 33]. OA is the most common skeletal condition indicative of degeneration from overuse or injury and is seen more commonly in lower-limb amputee populations [6, 22, 35]. Unilateral amputees show higher prevalence in knee and hip OA than able-bodied populations [11, 23, 24]. In addition, higher prevalence of knee OA occurs in the intact limb compared to the amputated side [43]. Increased joint degeneration stems from increased reliance on the intact limb, seen by higher power and joint reaction forces on the intact limb [12]. Lack of residual knee contribution may indicate socket restriction in addition to decreased load tolerance of residual limb tissues [7].

Hip Joint

The lack of a responsive ankle joint and reduced contribution of the residual knee places increased reliance on the ipsilateral hip joint while performing ADL [40]. The prevalence of hip joint OA is 44% on the residual limb compared to 15% on the intact limb and 11% in able-bodied populations [6, 36]. While the intact limb is a primary compensator for overall residual limb deficits, drastic increase in prevalence in OA indicate that greater stress is placed on the hip of the amputated limb [22].

Forces exerted on the residual limb hip joint are greater than those placed on any other lower limb joint during ambulation [22, 23]. Significantly less knee mobility and anterior center of mass indicates more vertical loading of the residual limb during heel contact through stance phase, supported by past reports of increased negative joint power seen in the hip and knee extensors, (Table 5). Furthermore, increased loading has been linked to kinematics associated with individuals predisposed to OA [6, 43].

In addition to compensations made at the hip in the sagittal plane, transverse and frontal plane kinematics indicate compensatory mechanisms similar to populations with “drop-foot,” a disorder in which the dorsiflexors do not function properly [26]. Adduction and external rotation are exhibited at the hip in an effort to swing the foot through to plant anterior to the body, mimicking a swing phase of gait [3].

Table 6: Mechanics associated with amputee gait according to previous research [11, 33, 40, 45].

		Kinematics	Kinetics	Gait Parameters
Knee	Residual	< Knee Flexion*	> Flexor and Extensor Negative work*	> Stance time*
		>Time to peak flexion	< Flexor/Extensor Positive work	< Step length*
		< Frontal ROM	> Duration of flexor/extensor work	< Swing time
			< VGRF	
	Intact		> VGRF*	> Step length
			> Extensor work	
Hip	Residual	> Flexion	> Positive and negative work, flexor/extensors*	> Trunk Flexion*
		< Extension*	> Negative work, frontal plane*	
		> Frontal ROM*		
		> External Rotation		
	Intact	> Adduction		

*Associated with OA predisposition [19, 23, 25, 43].

Silverman et al. [40] has suggested that positive hip joint power in the residual limb is a primary compensatory mechanism to initiate propulsion during ambulation. Additionally, hip and knee extensor strength should be prioritized to reduce loads placed on the joints of the residual limb, which coincides with major muscle groups correlated to moving at increased speeds [46]. Therefore, improving hip and knee extensor strength may improve one’s ability to ambulate effectively and execute ADL of increased difficulty [19, 46].

Stair and Directional Tasks

Stair descent is a difficult ADL for able-bodied and clinical populations. Individuals with a transtibial amputation more susceptible to falling exhibit altered gait parameters compared to transtibial amputees not susceptible to falling; however, both show compensatory mechanisms not present in able-bodied counterparts [45]. Amputees susceptible to falling selected higher ambulation speed, which suggests decreased stability according to past correlations between stability in clinical populations and walking speed [20]. These populations exhibit anterior trunk lean, possibly to compensate for decreased stability [20, 45]. Less stability in the residual limb leads to increased stance time and force exerted on the intact limb, which indicates limb asymmetry

[10, 33]. Work done by the hip extensors displays the most notable difference during stair ascent and descent as a controlling mechanism on the amputated limb [32].

Executing stair descent with a dynamic-response foot with a linear hydraulic system microprocessor knee allows amputees a more fluid gait cycle via steady limb loading and sagittal plane kinematics, which is similar to able-bodied persons, reducing the risk of development of OA [2, 25]. Regardless of the prosthetic device used for stair descent, researchers recommend more cautious strategies to protect the body during compensatory mechanics during strenuous tasks such as stair ascent and descent [45, 8].

Turning a corner or an unanticipated change in direction are other difficult ADL. Increased knee moments in the frontal and transverse planes are exhibited more so during an unanticipated 45 degree side step, increasing the likelihood of ACL injury [39]. To compensate for compromised knee stability, frontal plane hip ROM increases and researchers suggested that hip musculature is a limiting factor in completing a directional task [16]. Efforts to limit knee valgus, ground reaction force of landing and radius of the cutting direction may decrease ACL injury risk by decreasing uneven loading at the knee joint [21, 42]. Because the ACL helps stabilize the knee by limiting anterior tibial translation, compensatory efforts made by the knee extensors could provide a possible explanation for abnormalities as an effort to control movement of tibial translation [3].

The anterior cruciate ligament (ACL) is a structural component of the knee that aids in dynamic stability and is susceptible to injury during combined sagittal and frontal plane movements [3]. In previously- injured populations, stair descent indicates restricted knee flexion and knee moments, and more eccentric work performed by knee extensors compared to healthy counterparts [17]. Furthermore, ROM and joint work in all three planes of motion are significantly restricted during stair ascent and descent in adults with previously- injured ACL [11].

Increased hip abduction moment in the hip and knee are strong predictors for OA in able-bodied and amputee populations [12, 25]. Furthermore, strong hip adductor moments during gait have indicated a reduction in the risk of developing lower limb OA [19, 26]. Both of these variables are influenced by trunk and foot placement as a stabilizing mechanism, which is a possible indicator for the commonly seen hip abduction moments during amputee gait [16, 37]. Hip adductor moments occur when the force vector moves more medial to the body, and results in less force placed on the medial knee compartment [1, 16]. Efforts to condition the body to exhibit greater hip adduction moments may limit OA prevalence [19, 26].

Conclusion

Structural limitations of prosthetic devices do not allow amputee populations to execute tasks efficiently, which predisposes this population for secondary degenerative conditions such as OA. Restrictions of the residual limb caused by structural limitations of prostheses indicate that compensations occur elsewhere in the body, such as at the ipsilateral hip joint or the intact limb in unilateral amputees. Past research indicates that atypical mechanics are more prominent during ADL of greater difficulty than level-ground walking. Therefore, these mechanisms should be identified and corrected to reduce the increased prevalence of hip OA of the ipsilateral limb and knee OA of the contralateral limb. More specifically, research that identifies the relationship between the residual-limb knee joint and compensatory mechanisms during strenuous ADL directional tasks may improve quality of life in amputee populations by encouraging the design of prosthetic devices to represent efficient biomechanics while executing ADL.

CHAPTER 3: CONCLUSIONS AND RECOMMENDATIONS

CONCLUSION

The current study provided insight to the mechanical strategies used by unilateral, transtibial amputees to successfully execute strenuous ADL of stair descent and a non-linear directional task. Mechanics exhibited by amputees in the current study were similar to those reported in previous research. However, significant differences in forces, gait parameters and mechanics between residual, intact, and able-bodied participants were larger than previous reports of level-ground amputee gait.

Mechanics of the amputated limb did not indicate restriction of the knee joint from a prosthetic socket nor did results indicate compensatory movements in the residual hip joint. Previously reported reduced load tolerance of the residual limb indicated an intraindividual asymmetry due to compensatory mechanics that occurred in the intact limb [10, 11, 46]. These compensatory mechanics highlighted decreased hip adduction moments seen in amputees more so than able-bodied individuals, which may explain increased prevalence for OA in the intact limb.

Future research should extend the comparison of kinematic differences between amputees and able-bodied controls to the participants' whole body. The current study limited the mechanical comparison to the lower limbs, which did not quantify the use of handrails by the upper extremities or trunk position. Additionally, measuring the mechanics of the contralateral limb on the lowest step of the staircase may have provided a more complete understanding of the strategies that were implemented by amputee participants to execute strenuous ADL.

Populations with transtibial amputation may execute strenuous ADL with more variability compared to able-bodied populations. The able-bodied group employed mechanics that indicated less interlimb asymmetry and increased hip adduction moments that decreased likelihood for joint degeneration. If lower limb mechanics in amputees do not allow for efficient execution of ADL because of increased muscle work to maintain trunk and limb stability, increased conditioning of trunk and lower limb musculature may improve mechanics and delay joint degeneration.

LIMITATIONS

The biggest limitation of the current study was the variability between amputee participants. Activity level, time ambulating with a prosthesis, and body mass were all potential contributors to the high variance seen in most kinematic variables. This study recruited more amputee participants than the majority of research published in this field, which are often case study designed. Because variability in the mechanics seen in this population is high, comparison of results between studies within this field becomes difficult. Although the sample size in the current study was unlikely to highlight significant differences, providing exclusion criteria regarding amputee participant functionality in future research may reduce interindividual variability seen in the current study.

Although comparison between two tasks and three legs provided some insight to mechanics, variables measured in this study were limited to the lower limbs during the stance phase of one limb. Anecdotal evidence indicated that a higher percentage of amputee participants used the handrails than control participants. Measuring upper extremity and trunk mechanics during these tasks in future research may provide a more thorough explanation of the strategies amputee populations implement.

Analysis was restricted to a stance phase, which limits the comparison of mechanics. Future research comparing the mechanics in strenuous ADL tasks are encouraged to include the contralateral limb or include a full gait cycle of one limb.

PRACTICAL APPLICATIONS

Locating the cause of altered mechanics by examining whole-bodied mechanics during ADL execution in future research would better serve clinicians as a basis to correct detrimental mechanics by conditioning hip and knee musculature in the residual limb. Strengthening residual limb musculature may improve quality of life in amputee populations by increasing load tolerance in the residual limb, reducing compensatory asymmetries in the intact limb associated with OA.

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APPENDIX A

NORTHERN MICHIGAN UNIVERSITY

SCHOOL of HEALTH & HUMAN PERFORMANCE

CONSENT TO ACT AS A HUMAN SUBJECT

Subject Name (print):**_________________________________** Date **__________**

I hereby volunteer to participate as a subject in exercise and biomechanical testing. I understand that this testing is part of a study entitled: "Compensatory Hip and Knee Mechanics in Transtibial Amputees During Stair Descent and Directional Cutting Task". The purpose of the study is to investigate the compensations made at the hip joint during strenuous Activities of Daily Living tasks if and when the knee joint is restricted by a prosthetic socket.

1. I hereby authorize Mindie Clark, Sarah B. Clarke and/or assistants as may be selected by them to perform on me the following procedures:

    (a) I understand that I am being asked to descend a 4-step staircase at least 18 times during one session in the lab; practice trials to familiarize myself with the movement until I am comfortable are suggested.

    (b) Following stair descent, I will be asked to perform one of three directional tasks, starting with one foot and then the other: walking straight ahead two gait cycles (4 steps), 45-degree left side- step or cross-over, or 45-degree right side- step or cross-over. All of these directions will be performed in a random order. These tasks will be demonstrated to me prior and at least familiarization trials for each direction will be performed as specified above.

    (c) I understand that markers will be placed on my hip, both legs, feet and/or prosthetic device, throughout tasks specified in (a-b) to measure movement and angles of my lower limb.

2. The procedures outlined in paragraph 1 [above] have been explained to me.

    I understand that the procedures described in paragraph 1 (above) involve the following risks and discomforts: temporary muscle pain and soreness due to physical activity may be expected. Minor skin irritation due to tape used to secure reflective markers may occur. Additionally, falling and ankle sprain risk during stair descent and directional tasks are present. I understand that handrails are secured in place for me to use at my discretion to aid in stability.

I understand that I can terminate any test at any time at my discretion. Moreover, I should cease any test if I experience any abnormalities such as dizziness, light-headedness, or shortness of breath, or abnormal pain, numbness or tingling, etc.

3. I have been advised that the following benefits will be derived from my participation in this study: aside from the educational benefit of learning of task analysis, there are no direct benefits to me.
4. I understand that Mindie Clark, Sarah B. Clarke and/or designated assistants will answer any questions or concerns that I may have at any time concerning these procedures and/or investigations.
5. I understand that all data concerning myself will be kept confidential and available only upon my written request. I further understand that in the event of publication, no association will be made between the reported data and myself.
6. I understand that there is no monetary compensation for my participation in this study. I may request Mindie Clark or Sarah B. Clarke send the results to me once they are made available to the public.
7. I understand that in the event of physical injury directly resulting from participation, compensation cannot be provided. However, if injury occurs, emergency first aid will be provided and the EMS system activated.
8. I understand that I may terminate participation in this study at any time without prejudice to future care or any possible reimbursement of expenses, compensation, or employment status.
9. I understand that if I have any further questions regarding my rights as a participant in a research project I may contact Dr. Robert Winn (906-227-2300) rwinn@nmu.edu, Assistant Provost of Graduate Education/Research of Northern Michigan University. Any questions I have regarding the nature of this research project will be answered by Mindie Clark (530-375-0768) or minclark@nmu.edu.

Subject's Signature:____

Witness:_____________________________________ Date:________

APPENDIX B

COMPENSATORY HIP AND KNEE MECHANICS IN TRANSTIBIAL AMPUTEES DURING STAIR DESCENT AND DIRECTIONAL TASK

General Information

Name: Age: Ht: Wt:

Amputee N/A R L Length of residual limb:

Age Amputated:

Reason Amputated:

Prosthetic Device currently being used:

Length of time using current device:

Which leg do you normally kick with (before amputation): R L Not sure

Do you feel numbness or tingling in the amputated limb during walking or physical activity?

Y N

If yes, do these sensations impair your ability to perform daily tasks or physical activity?

Please describe severity and location of sensation during specific activities

Please take the ADL survey on the page of this page

Research Assistant Use:

Marker Set: Control : R L Amputee: R L

Cadence Pace: __________

Familiarization: _________

Order		✓ ✓ ✓
___	L Straight Ahead	___ ___ ___
___	R Straight Ahead	___ ___ ___
___	L Crossover	___ ___ ___
___	R Crossover	___ ___ ___
___	L Side Step	___ ___ ___
___	R Side Step	___ ___ ___

Activities of Daily Living level of Independence Survey

Please rank the following tasks on competency of independent functionality: 1-4. If you use a prosthetic, please indicate with Y/N if you do or do not use your device to complete the following task

1: Cannot complete

2: Needs assistance from outside source

3: Able to perform with limited help

4: Able to perform at ease

Bathing:	1_____	2_____	3_____	4_____	N/A: _____
Dressing:	1_____	2_____	3_____	4_____	N/A: _____
Grooming:	1_____	2_____	3_____	4_____	N/A: _____
Toileting:	1_____	2_____	3_____	4_____	N/A: _____
Moving from bed/chair:	1_____	2_____	3_____	4_____	N/A: _____
Walking:	1_____	2_____	3_____	4_____	N/A: _____
Climbing Stairs:	1_____	2_____	3_____	4_____	N/A: _____
Shopping:	1_____	2_____	3_____	4_____	N/A: _____
Cooking:	1_____	2_____	3_____	4_____	N/A: _____
Housework:	1_____	2_____	3_____	4_____	N/A: _____
Laundry:	1_____	2_____	3_____	4_____	N/A: _____
Driving:	1_____	2_____	3_____	4_____	N/A: _____

If the results of this study is made available to the public, would you like an email by Mindie regarding this information?

Y N

MEMORANDUM

TO: Mindie Clark
School of Health and Human Performance

CC: Sarah Clarke
School of Health and Human
Performance FROM: Robert Winn, Ph.D.
Interim Dean of Arts and Sciences/IRB Administrator

DATE: December 20, 2017

RE: Modification to HS17-908
Original IRB Approval Date: 12/4/2017
Modification Approval Date: 12/20/2017
"Compensatory Hip and Knee Mechanics in Transtibial Amputees During Stair Descent and Directional Cutting Tasks"
Your modification for the project "Compensatory Hip and Knee Mechanics in Transtibial Amputees During Stair Descent and Directional Cutting Tasks" has been approved under the administrative review process. Please include your proposal number (HS17-908) on all research materials and on any correspondence regarding this project.

Any additional changes or revisions to your approved research plan must be approved by the IRB prior to implementation. Unless specified otherwise, all previous requirements included in your original approval notice remain in effect.

If you complete your project within 12 months from the date of your approval notification, you must submit a Project Completion Form for Research Involving Human Subjects. If you do not complete your project within 12 months from the date of your approval notification, you must submit a Project Renewal Form for Research Involving Human Subjects. You may apply for a one-year project renewal up to four times.

NOTE: Failure to submit a Project Completion Form or Project Renewal Form within 12 months from the date of your approval notification will result in a suspension of Human Subjects Research privileges for all investigators listed on the application until the form is submitted and approved.

If you have any questions, please contact the IRB at hsrr@nmu.edu.