Mobility and Agility During Locomotion in the Mark III Space Suit

Conor R. Cullinane; Richard A. Rhodes; Leia A. Stirling

INTRODUCTION: The Mark III (MIII) space suit assembly (SSAs) implements a multibearing, hard-material hip brief assembly (HBA). We hypothesize that: 1) the MIII HBA restricts operator mobility and agility which manifests in effects to gait parameters; 2) the waist bearing provides rotational motion, partially alleviating the restrictions; and 3) there are resistive, speed-dependent torques associated with the spinning bearings which further diminish mobility and agility.

METHODS: A subject (Suited and Unsuited) performed two planetary tasks—walking forward (WF) and backward (WB). An analysis of variance (ANOVA) and post hoc comparisons were performed to determine interaction effects. Motion capture data was processed to obtain gait parameters: static base (m), dynamic base (m), step length (m), stride length (m), cadence (steps/min), center of mass speed ($m \cdot s^{-1}$), foot clearance (toe and heel) (m), and bearing angular velocities ($° \cdot s^{-1}$).

RESULTS: The static base when Suited (0.355 m) was larger than Unsuited (0.263 m). The Suited dynamic base (pooled, 0.200 m) was larger than both Unsuited WF (0.081 m) and WB (0.107 m). When Suited, the operator had lower clearance heights. The waist bearings provided about 7.2° of rotation when WB and WF. The maximum torque, while WF, in the right upper and mid bearings was 15.6 ± 1.35 Nm and 16.3 ± 1.28 Nm.

DISCUSSION: This study integrated suit component properties and the emergent biomechanics of the operator to investigate how biomechanics are affected. The human hip has three collocated degrees of freedom (DOFs), whereas the HBA has a single DOF per bearing. The results can inform requirements for future SSA and other wearable system designs and evaluations.

KEYWORDS: programmed motion, static base, dynamic base, step length, stride length, clearance analysis, bearing analysis, joint torque, degrees of freedom, biomechanics, anthropometry, computer-aided design, motion capture.

Cullinane CR, Rhodes RA, Stirling LA. *Mobility and agility during locomotion in the Mark III space suit*. Aerosp Med Hum Perform. 2017; 88(6):589–596.

Manned spaceflight necessitates life support for crewmembers during extravehicular activities (EVAs) in the form of space suit assemblies (SSAs), which should maximize human performance and efficiency while preventing injury.[7] New suit designs consider the interaction of mass, volume, walking effort, mobility, agility, and suit fit.[2] By increasing mobility and reducing deviations from unsuited operator kinematics, injury risk and metabolic cost may be reduced,[22] and could extend exploration capabilities and improve exploration mission operations. While more recent SSA designs increase mobility, these systems lead operators to perform predetermined (programmed) motions that are inconsistent with natural biomechanics, causing the operator to fight against the suit to achieve mission goals.[13] Crewmembers spend many hours in a given suit to learn and adapt to the programmed motions, with training important for learning to operate a suit and discover ways to map the human joint degrees of freedom (DOFs) with the suit DOFs.[6]

The Mark III (MIII) Planetary SSA was developed to evaluate technology that extends planetary explorer capabilities beyond those of the Apollo A7L, the suit used for lunar exploration. One technology tested is a hip brief assembly (HBA) that implements a multibearing hard-material (IM7–997–3 and S-Glass, improved strength E-fiberglass)[3,16] solution to provide constant-volume joints. The Mark III HBA architecture implements three bearings in series for each hip, an upper, mid, and

From Health Sciences and Technology, and the Institute for Medical Engineering and Science, Aeronautics and Astronautics, Massachusetts Institute of Technology, Cambridge, MA; Harvard Medical School, Boston, MA; and Advanced Pressure Garment Technology Development Lab, Crew Survival and Space Suit Systems Branch EC5, NASA Johnson Space Center, Houston, TX.

This manuscript was received for review in April 2016. It was accepted for publication in March 2017.

Address correspondence to: Conor R. Cullinane, Health Sciences and Technology, Massachusetts Institute of Technology, 77 Massachusetts Ave., Cambridge, MA 02139; cullincr@mit.edu.

DOI: https://doi.org/10.3357/AMHP.4650.2017

rolling convolute bearing (**Fig. 1**), where the upper bearing is the most proximal and the rolling convolute bearing is the most distal. The upper and mid bearings provide hip flexion/extension and adduction/ abduction. Cowley et al.[4] computationally estimated the HBA hip rotation to have 84° of flexion, 7° of pure abduction, and 93° of transversely rotated abduction (a programmed motion that combines hip flexion with abduction).

Experiments at NASA comparing the MIII HBA to the Apollo A7L hip joint reveal the HBA to have improved performance over its predecessor in two important categories: walking effort and retrieving a rock sample. However, the HBA has limitations in bent torso stability, standing on one knee, and requires programmed motion.[2] While the HBA eliminates the volume change during joint motions, it introduces an instability that could lead to altered biomechanics during planetary tasks.[9,13] Operators wearing earlier EVA suits have developed a range of injuries in a range of locations, including the shoulder and hip, due to prolonged use, including erythema, abrasions, muscle soreness/fatigue, paresthesia, bruising, blanching, and edema.[14,17] If the suit alters natural biomechanics, it may result in increased injury risk over long-term usage inherent in locomotion and utility tasks.[15] Even a small alteration in natural biomechanics can result in injury when amplified through many cycles.[11,13]

There have been many studies in the biomechanics literature examining and characterizing unsuited gait. It has been shown that cadence (the number of steps per unit time) increases while step length and speed decrease when walking backward (WB) as compared to walking forward (WF).[8,20] The gait cycle (1/cadence) and stride length are also functions of speed and direction (WF or WB).[8] Changes in dynamic base (the perpendicular distance between heels during double stance) can provide insight on overall stability, with a wide dynamic base providing improved balance and stability for controlling body mass over the supporting limb. In child development, the dynamic base decreases until the age of three and a half,[19] which can be interpreted as an improvement in coordination and balance, causing the child to require less stability through foot placement while controlling body mass. Within the elderly population, the dynamic base increases among those at risk of falling and when walking at faster speeds,[12] providing additional stability. LaFiandara et al.[10] found adding a load to the body decreased the transverse pelvic rotation and stride length while maintaining speed and increasing cadence. Abe et al.[1] postulate that the change in stride length and cadence may be due to attempts to minimize energy expenditure.

Fig. 1. The Mark III HBA.

In addition to adding mass, the spacesuit provides resistive forces to the operator. Cullinane et al.[5] previously characterized the torque required to rotate the HBA upper and mid bearings for a range of angular velocities. Coupling these data with the measured bearing angular velocities during ambulation may lead to an improved understanding of the underlying programmed motions and provide future design guidelines for wearable systems. In this paper, we present the results of a pilot study of locomotion while Unsuited and Suited with the MIII SSA. This study aims to integrate underlying suit component characteristics with the emergent biomechanics of the operator to investigate how biomechanics are affected by the MIII, specifically how the HBA architecture effects hip kinematics and dynamics. We hypothesized that 1) the MIII HBA architecture has DOF limitations that restrict operator mobility and agility. The limitations manifest in effects to both static and dynamic (WF and WB) gait parameters. 2) Based on subjective feedback from experienced suit testers, the waist bearing provides rotational motion in the transverse plane during ambulation, partially alleviating mobility restrictions introduced by the HBA. 3) Although the HBA volume does not change during hip joint motion, there is still a resistive speed-dependent torque associated with the spinning bearings. Those torques further diminish the mobility and agility of the operator, requiring increased hip joint torques along the limited DOFs. The data presented in this study are relevant for improving future SSA engineering through design requirements development and evaluation methods.

METHODS

Subject

This pilot study was performed on a single subject who represents an astronaut. The subject was 30 yr old and 72″ tall, which falls within the astronaut selection criteria for age (26–46 yr old) and height (62–75″) of astronaut candidates. The subject had extensive experience operating the suit, enabling the assumption of a trained operator. The subject was cleared with a class I medical to participate as a suit operator. The study protocol was approved by the NASA Johnson Space Center IRB and the subject provided written informed consent prior to participation.

Equipment

The study was performed in the Anthropometrics and Biomechanics facility (ABF) at the NASA Johnson Space Center. Data were collected using 11 Vicon Bonita cameras imaging at 100 frames per second. The MIII was pressurized to nominal suit pressure (4.3 psi) in a tethered configuration [i.e., not with the closed-loop portable life support system (PLSS)]. The MIII in the tethered configuration weighs 59 kg.[2,21] No mass was added in this configuration to simulate the PLSS. While unsuited, the subject wore a compression shirt and pants.

Procedure

For the Unsuited condition, passive reflective motion capture markers were placed at anatomical landmarks (**Fig. 2**, top). The subject stood still to obtain a static pose, then performed trials that included WF (12 trials) and WB (6 trials) within the motion capture volume (10 m long by 1 m wide walkway). Following Unsuited operations, the subject donned the MIII SSA, which was then pressurized. For the Suited configuration, the motion capture markers were placed external to the SSA (Fig. 2, bottom). It can be difficult, or sometimes impossible, to identify anatomical landmarks to guide marker placement; rather the placement was intended to highlight features of interest on the suit, especially on the HBA. While Suited, the static pose, as well as WF (10 trials) and WB (5 trials) were repeated.

Data Analysis

Logarithmic and hyperbolic fit equations from Grasso et al.[8] were used to predict the Unsuited stride length and cadence, respectively. The hyperbolic equation was originally expressed for gait cycle. Here we take the inverse to predict cadence. An HBA model was created in the computer-aided design (CAD) package SolidWorks[18] (Fig. 1) to predict functional motion envelopes.[4] The position data (x, y, z) of the motion capture markers were used to calculate static and dynamic gait parameters for the subject while Suited and Unsuited. For the dynamic trials, the data sets were trimmed to include the steady state gait within the capture volume, which tended to include 2–3 strides.

The gait parameters examined for Hypothesis 1 were static base (m), dynamic base (m), step length (m), stride length (m), cadence (steps/min), center of mass speed ($m \cdot s^{-1}$), and foot clearance (toe and heel, m). The static base was defined as the distance between the left and right heel markers when the subject was standing in a static pose. The HBA CAD model was also used to predict the minimum static base while suited by orienting the HBA to minimize the distance between the rolling convolute joints. Step length was defined as the parallel distance between two consecutive heel strikes of opposite feet, while stride length was defined as the distance between two consecutive heel strikes of the same foot. Cadence was defined as the step length divided by the time to complete that step. Speed was determined as a center of mass speed and was calculated from a chest or hard upper torso (HUT) marker. The heel and toe clearance was defined as the vertical distance above the calibrated origin, where the calibrated origin was the height when the foot was in contact with the ground.

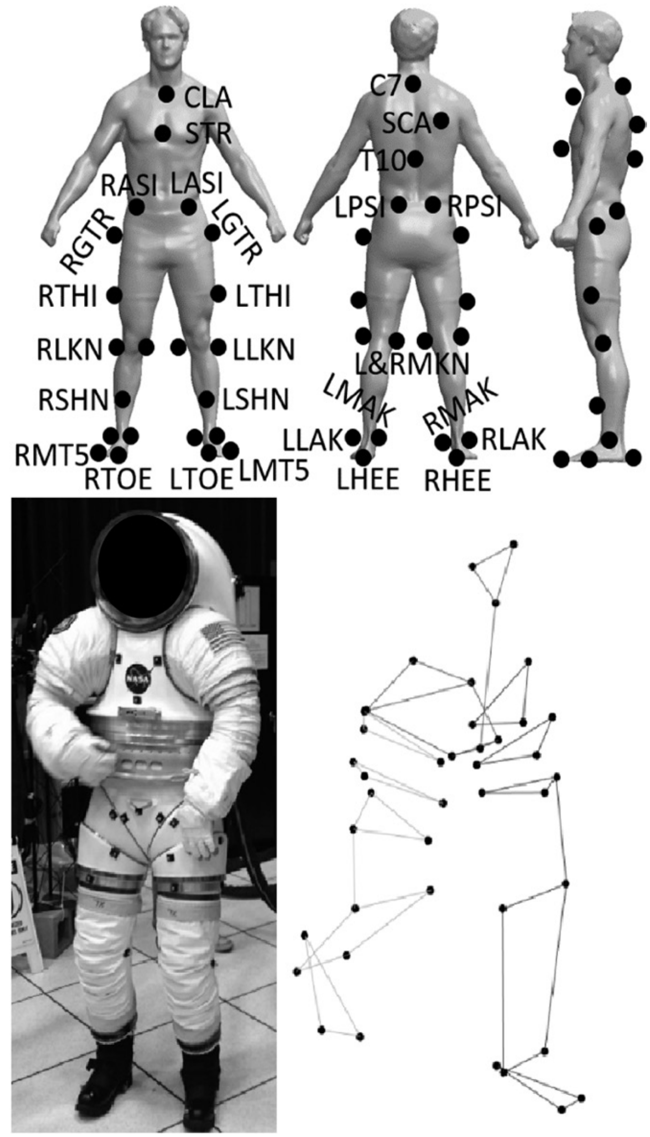

Fig. 2. Unsuited (top) and Suited (bottom) marker placement. "L" and "R" denote left and right. Markers were placed on the Unsuited subject at the clavicle (CLA), sternum (STR), anterior superior iliac spines (ASI), greater trochanters (GTR), thigh (THI), medial knees (MKN), lateral knees (LKN), shins (SHN), medial ankle (MAK), lateral ankle (LAK), 5th metatarsals (MT5), big toes (TOE), heels (HEE), posterior superior iliac spines (PSI), 10th thoracic vertebrae (T10), 7th cervical vertebrae (C7), and scapula (SCA). When Suited, three markers were evenly spaced on each of the bearings. Two triads of markers were placed on the main brief section, one in the front and one in the rear. Another triad was placed on the hard upper torso (HUT). Other markers were chosen to replicate anatomical marker placements (MKN, LKN, MAK LAK, MT5, TOE, and HEE).

The parameter necessary for testing Hypothesis 2 was the waist bearing rotation (degrees), which was defined as the difference in motion between the HUT and main brief section. Normal vectors were calculated from the motion capture marker triads on both the front of the HUT and HBA. For example, if the three markers in the triad were labeled A, B, and C, the normal vector could be calculated as $N = \overrightarrow{AB}\, X\, \overrightarrow{AC}$. With the two normal vectors (N1 and N2) calculated from each triad, the angle between the vectors was calculated as shown in Eq. 1. The difference between the maximum and minimum angles for each trial is the range that the waist bearing rotated during ambulation.

$$\theta = \cos^{-1}\left(\frac{N1 \cdot N2}{N1N2}\right) \quad \text{Eq. 1}$$

To test Hypothesis 3, the upper/mid bearing angular velocities ($° \cdot s^{-1}$) are determined. The bearing angular velocities, $(\omega_x, \omega_y, \omega_z)$, were calculated with the system of equations:

$$\frac{x_1(t+\Delta t)-x_1(t)}{\Delta t} - \frac{\bar{x}(t+\Delta t)-\bar{x}(t)}{\Delta t} = \omega_y(z_1-\bar{z}) - \omega_z(y_1-\bar{y}) \quad \text{Eq. 2}$$

$$-\left(\frac{y_1(t+\Delta t)-y_1(t)}{\Delta t} - \frac{\bar{y}(t+\Delta t)-\bar{y}(t)}{\Delta t}\right) = \omega_x(z_1-\bar{z}) - \omega_z(y_1-\bar{y}) \quad \text{Eq. 3}$$

$$\frac{z_1(t+\Delta t)-z_1(t)}{\Delta t} - \frac{\bar{z}(t+\Delta t)-\bar{z}(t)}{\Delta t} = \omega_x(z_1-\bar{z}) - \omega_y(y_1-\bar{y}) \quad \text{Eq. 4}$$

where (x_1, y_1, z_1) is the position of a single marker on a bearing and $(\bar{x}, \bar{y}, \bar{z})$ is the average of all three position markers, the centroid. Here, ω_x and ω_y represent angular velocities that result in pivoting out of plane, while ω_z is the bearing rotation and results in a change in the HBA shape. The estimated bearing torques to achieve the calculated angular velocities are then calculated using the parabolic fits from Cullinane et al.[5]

Statistical Analysis

A 95% confidence interval for the cadence and step length were constructed to assess the predicted values from Grasso et al.[8] An analysis of variance (ANOVA) was performed for each gait parameter to examine effects of task type (WF, WB) and suit configuration (Suited, Unsuited). Post hoc comparisons were performed within each gait parameter between task types and suit configurations. Two-sample *t*-tests were used to compare between left/right leg and Unsuited/Suited heel and toe height within the clearance analysis. Two sample *t*-tests were also used to compare the bearing rotation and required torques between the upper and mid bearings on both the left and right side. A one-sample *t*-test was used to test the null hypothesis that the mid bearing data when WB comes from a normal distribution with mean equal to zero. An outlier analysis was performed to give statistical reasoning for removing any data points. Significance was set at $P < 0.05$ for all tests.

RESULTS

As this was a pilot study that contained only one subject, Unsuited stride length and cadence were compared with values estimated using the regressions from Grasso et al.[8] The measured mean speed, taken from the speed of a chest marker, for WF ($1.12\ m \cdot s^{-1}$) and WB ($0.904\ m \cdot s^{-1}$) provided predicted stride lengths of 1.30 m and 1.19 m for WF and WB, respectively. These values were within the 95% confidence intervals of the stride lengths determined from the measured data (1.31 ± 0.0262 m and 1.20 ± 0.0347 m). The predicted cadence for WF and WB (102 steps/min and 94.4 steps/min) were outside the 95% confidence intervals (98.6 ± 1.58 steps/min and 79.2 ± 2.40 steps/min).

The static base for Unsuited and Suited was 0.263 m and 0.355 m, respectively. Using the HBA CAD model, the sagittal plane vertical distance from the rolling convolute medial edge to the center of the main brief section (hip joint center) was calculated to be 218.9 mm. The transverse plane horizontal distance from the rolling convolute medial edge to the center of the HBA was calculated to be 41.16 mm. The minimum angle created by the brief at the hip joint was 10.65°. The subject's leg length from the greater trochanter (GTR, Fig. 2) to the floor was 0.953 m. Therefore, with the legs pressed against the medial edge of the rolling convolute joint, the predicted static base was 0.354 m.

When analyzing the dynamic gait parameters, the outlier analysis found a single outlier for the Suited WF dynamic base (0.64 m), which may have been the result of an abnormal step (i.e., stepping with a wider base than normal to catch oneself when off balance). This point was removed from the analysis as it was not considered a steady state step. An ANOVA found a significant interaction effect of task (WF/WB) and configuration (Suited/Unsuited) for all gait parameters ($P < 0.005$) (**Table I**). When contrasting the treatment conditions, there was a statistically significant difference between Suited and Unsuited for WF and WB found for all gait parameters ($P < 0.05$), except for the mean dynamic base and cadence between the Suited WF and WB conditions (Table I). There was no statistical difference between WF and WB ($P = 0.069$). The mean waist bearing rotation ranges were 7.18° and 7.23° for WF and WB, respectively, which were not significantly different ($P = 0.928$).

The clearance analysis results are shown in **Fig. 3**. The ANOVA results showed no effect of Leg (Right vs. Left) ($P = 0.161$) on the maximum clearance height. However, there was an effect of anatomical landmark (Heel vs. Toe), suit configuration (Suited vs. Unsuited), and task (WF vs. WB) ($P < 0.005$).

Table I. Mean and SD of Dynamic Gait Parameters.

	MEAN STEP LENGTH (m)	STEP LENGTH SD (m)	MEAN STRIDE LENGTH (m)	STRIDE LENGTH SD (m)	MEAN CADENCE (STEPS/MIN)	CADENCE SD (STEPS/MIN)	MEAN SPEED ($m \cdot s^{-1}$)	SPEED SD ($m \cdot s^{-1}$)	MEAN DYNAMIC BASE (m)	DYNAMIC BASE SD (m)
Suited WF	0.595	0.030	1.243	0.053	*88.794	3.017	1.061	0.058	†0.190	0.027
Unsuited WF	0.631	0.045	1.309	0.050	98.651	4.262	1.117	0.076	0.081	0.021
Suited WB	0.476	0.044	0.860	0.085	*90.181	5.477	0.833	0.082	†0.220	0.037
Unsuited WB	0.549	0.070	1.197	0.056	79.160	5.190	0.904	0.096	0.107	0.032

The *mean dynamic base and the †mean cadence are not statistically different between WF and WB while suited.

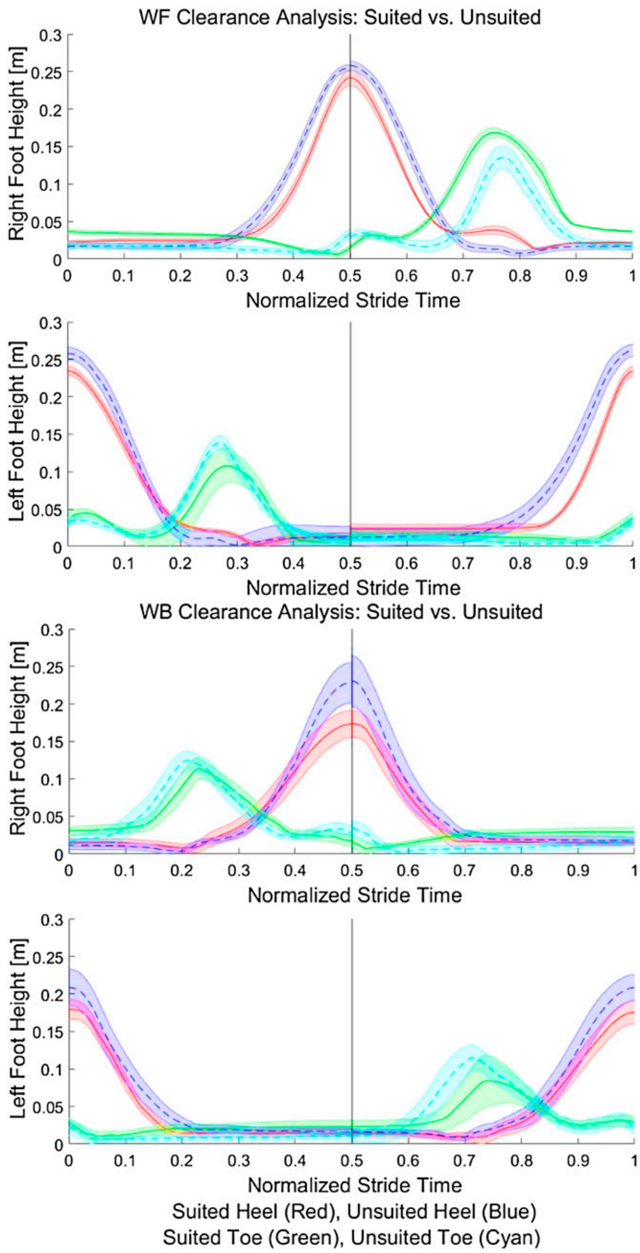

Fig. 3. Clearance analysis. The two top plots are the heel and toe clearances while WF and the two bottom plots are while WB. The dashed lines represent Unsuited and the solid lines represent Suited.

Within each suit configuration and task combination, the maximum heel clearance was greater than the toe clearance ($P < 0.005$). For both WF and WB, the maximum clearance for the Unsuited Heel and Toe was larger than when Suited ($P < 0.005$).

The upper and mid bearing kinematics and torque estimates are shown in **Fig. 4**. When WF, the upper and mid hip bearings rotated in opposite directions simultaneously, allowing the HBA to change shape and follow the subject's limb. However, the maximum rotation angle for the upper bearings were larger than the mid bearing ($P < 0.005$). When WB, the upper bearings rotated; however, the mid bearing had very little motion. For example, the maximum rotation in the right upper (23.1 ± 3.97°) and mid (2.11 ± 2.93°) bearing during the right swing phase while WB were different ($P < 0.005$). The mean mid bearing torque when WB was not different from zero rotation ($P = 0.111$). The angular velocities and torques are consistent with this motion profile. For example, when WF, the maximum torque in the right upper (15.6 ± 1.35 Nm) and mid (16.3 ± 1.28 Nm) bearings were not different ($P = 0.1729$).

DISCUSSION

This study aimed to integrate the underlying suit component properties to the emergent biomechanics of the operator to describe how the biomechanics were affected by the MIII. Specifically, we hypothesized that 1) the MIII HBA architecture had DOF limitations that restricted operator mobility and agility (as defined by static and dynamic gait parameters); 2) the waist bearing provided rotational motion in the transverse plane during ambulation, partially alleviating mobility restrictions introduced by the HBA; and 3) there was a resistive speed-dependent torque associated with the spinning bearings that further diminished the mobility and agility of the operator, requiring increased hip joint torques along the limited rotational DOFs.

Prior to examining the hypotheses, the Unsuited pilot data were first compared to the literature. Although the stride lengths were predicted accurately for the experimental walking speed, the cadences were slower than predicted. Also, this subject had a lower Unsuited cadence and stride length when WB than WF, which is not consistent with previous studies that show an increase in cadence when WB to compensate for the decrease in stride length.[8,20] Here we observed that when Unsuited while WB, the subject had a wider dynamic base, smaller stride length, smaller step length, slower cadence, and slower speed, which all provide improved control of body mass over the supporting limb.[19] The selection of increased stability may be due to the experimental setup, which used an elevated walkway. The walkway may be easier to navigate when WF with visual cues, but more difficult when WB because it was narrow (1 m). This subject may have been more cautious than seen in the literature.

When Suited, an operator is not only manipulating the suit, but also carrying its weight. As mentioned previously, carrying a load on top of body mass decreases stride length while increasing cadence.[10] Our data show a decrease in stride length when Suited; however, the change in cadence is not consistent. When WB and WF, the cadence increases and decreases, respectively, from the Unsuited to Suited condition. If the weight alone was altering gait, we would expect consistent change across WF and WB. This implies that the differences in the Unsuited and Suited gait may have additional factors.

The heel clearance analysis showed the Suited condition resulted in lower clearance heights, which could be due to motion restrictions, or could be related to adapting the motion to minimize added torques due to the suit. The reduction in the maximum clearance heights of both heel and toe, for both WF and WB, confirm that operator mobility and agility is being

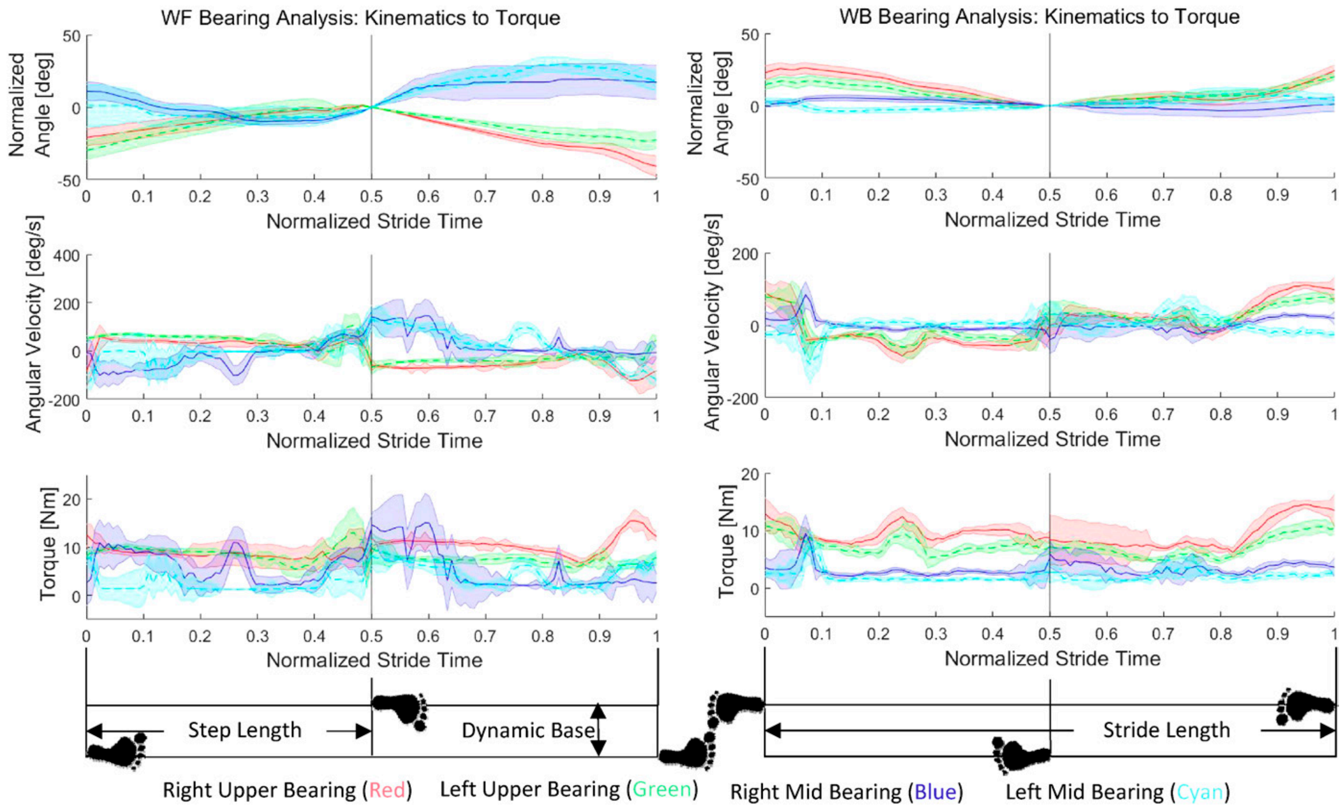

Fig. 4. HBA bearing analysis. The left side of the figure represents WF data, while the right represents the WB data. The top plot on both sides represents the rotational displacement angle normalized to be zero during double stance (x = 0.5 on the x axis). The middle plots represent the calculated angular velocities that achieve the experimental displacements in the top plots. Finally, the bottom plots show the calculated torques required to achieve the angular velocities. The bearings, matched to the legend, are seen in Figure 1. In each plot, the solid lines represent data from the right bearings and the dashed lines represent the left bearings.

restricted by the suit. The diminished mobility and agility, when compared to Unsuited, also manifested as a decrease in speed, stride length, and step length for both WF and WB. In addition to the added weight, these changes may be a result of the interaction and disparity between the natural hip biomechanics with the operational motion envelope of the HBA. The MIII HBA has a smaller range of motion than the human hip, as well as fewer DOFs. A human hip joint has three DOFs located within the single joint, whereas the HBA has a single degree of freedom per bearing and those are spread out distally from the hip. This is highlighted in the comparison of the Suited and Unsuited static bases.

Since a difference in static base exists, we examined if the architecture of the suit contributed to the wider base. The CAD HBA model confirmed the limitations in adduction and showed that the selected Suited static base was essentially the same as the minimum adduction permitted by the architecture. In this configuration, the medial thigh of the Suited operator contacted the medial edge of the rolling convolute joint, which acted as a physical hard-stop. This architecture also restricted the dynamic base, but Hypothesis 2 explores how the waist bearing may provide additional motion to partially alleviate that restriction during dynamic task performance.

When Unsuited, the dynamic base was larger when WB than WF. This finding is different than the Suited dynamic base (pooled, 0.200 m), which was found to be larger than both Unsuited WF (0.081 m) and WB (0.107 m). This is consistent with the HBA limiting adduction as the minimum dynamic base was seen to have an adduction hard-stop while Suited that forced the dynamic base to be larger than the naturally selected dynamic base observed in the Unsuited case.

In the CAD model, when manipulating the HBA into the double stance gait phase configuration, the dynamic base was predicted to increase (0.626 m) when no other degrees of freedom were altered. However, this increase in dynamic base was not seen in the measured data. During ambulation, there was actually a reduction in base as compared to the static condition. We hypothesized that this could be attributed to the waist bearing, which rotates to allow further adduction of the compound hip assembly. The CAD model and static base showed that the HBA was in its most adducted position when in the static pose. When walking, the HBA dynamically changes configuration as the operator drives the rotation of the bearings. Any rotation in the upper and mid bearings away from their static pose locations decreases the adduction of the operator due to the irregular geometry of the HBA sections between the bearings. This study confirmed Hypothesis 2 as a nonzero waist bearing rotation was measured, which allowed the operator to have foot placement closer to the walking centerline and reduced the dynamic base. The waist bearing connects the HUT to the hips, transferring load while allowing for rotation along the bearing (Fig. 1). Here we found that the waist bearing rotations were not statistically different in the WF and WB tasks. Thus, it is consistent that the dynamic bases were not significantly different when WF and WB. The HBA design increased static base width, but the waist bearing provides transverse plane rotation to reduce that effect during dynamic motion. This is similar to the Unsuited case, where the rotation in the spine provides this same reduction in static v. dynamic base.[23]

Hypothesis 3 was also supported, as the circumferential rotation of the bearings during Suited WF and WB created additional torques for the user. The motion of the bearings is not aligned with the natural desired joint trajectories because of DOF limitations inherent in the architecture, which produces programmed movements. The operator drives the HBA configuration through limb contact with the interior of the HBA as the hip is rotated. The limb, however, has to traverse the trajectories that the bearings and HBA geometry will allow, while exerting additional required torques to overcome the resistive bearing torques. To generate bearing rotation, the force has to be transferred from the operator's limb to the SSA distal to the bearings, increasing the moment arm of the applied force, and resulting in an increase in the required joint torques. As seen in the estimated torques, there is a difference between the bearing performance during WF and WB. When WF, the upper and mid bearings rotate in opposite directions. While WB, the upper bearing rotates while the mid bearing only rotates slightly or not at all. Thus, when WF, complementary motion occurs in both bearings, while when WB, only the upper bearing is necessary to achieve the foot placement observed. This is consistent with the strategy measured through the dynamic gait parameters. The smaller step lengths when WB as compared to WF while Suited are associated with less hip extension (as compared to the greater hip flexion when WF) and thus less bearing rotation. This strategy may be an attempt to minimize energy expenditure as proposed by Abe et al.[1] When WB, the overall speed and step length decrease from the Unsuited to Suited configuration, while the cadence increased. This shows that when WB, it may be more efficient to only force one bearing to spin and take smaller faster steps than it is to spin both bearings and take larger, slower steps while fighting the greater amount of resistive torque.

This pilot study was limited in that it only included one subject. Thus, conclusions and trends must be considered appropriately. The data show sufficient evidence to support the hypotheses and can be used to inform future studies with an increased sample size. An experimental limitation was the total distance that a subject could walk. It would be beneficial to have a larger track to walk, providing more continuous strides per trial. It may also be beneficial to remove the elevated walkway and determine if that was truly the root cause of the need for increased control and stability when Unsuited and WB. In this study, the spacesuit added weight to the subject and we examined the combined effect of the weight with the system architecture. However, the PLSS weight was not included and future studies should consider that added effects of the PLSS such as the changes to the center of gravity and total load can affect gait parameters. Also, future studies could examine the addition of comparable weight in the Unsuited configuration applied at a similar center of gravity locations without limiting mobility to control for load effects on mobility restriction. To determine the comparable weight, a differentiation should be made between the load transferred to the ground through the operator and the load transferred to the ground through the suit. This could be further examined by comparing total ground reaction forces (GRFs) with force plates and the GRFs on the foot using foot sole pressure sensors.

The results of this study are beneficial in informing and evaluating future SSA design requirements, as well as those for other wearable systems. For example, the results of Hypothesis 1 showed altered gait because of the SSA architecture. Design requirements on certain static and dynamic gait parameters (e.g., step length or dynamic base) could be created to inform maximum allowable percent deviations from nominal kinematics. For example, a requirement could be imposed that the dynamic base must permit a certain tolerance from the Unsuited mean. Requirements can also be generated for specific suit components, such as the HBA bearings or waist bearings, requiring a minimum performance, or a maximum allowable added torque for relevant angular velocities. Future studies can also inform design requirements based on energy expenditure differences between the Unsuited and Suited condition, supplementing the biomechanical analyses with metabolic analyses.

This study aimed to integrate the underlying suit component properties to the emergent biomechanics of the operator to describe how an operator's biomechanics are affected by the MIII. Specifically, we showed that 1) the MIII HBA architecture had DOF limitations that restricted operator mobility and agility. The limitations manifested in effects to both static and dynamic gait parameters. 2) The waist bearing provided rotational motion in the transverse plane during ambulation, partially alleviating mobility restrictions introduced by the HBA. 3) Although the HBA volume does not change during hip joint motion, the resistive speed-dependent torques associated with spinning the bearings further diminish the mobility and agility of the operator, requiring increased hip joint torques along the limited rotational DOFs and reducing ground clearance.

ACKNOWLEDGMENTS

This work was supported by NASA award NNX15AR20G and by the National Space Biomedical Research Institute through NASA NCC 9-58. The views expressed in this paper are not endorsed by the sponsors. The authors would like to thank Alan Natapoff from MIT, and Matthew Cowley, Scott England, Dan Nguyen, and Robert Sweet from the ABF.

Authors and affiliations: Conor Cullinane, B.S., Health Sciences and Technology, Aeronautics and Astronautics, Massachusetts Institute of Technology, Cambridge, MA, and Harvard Medical School, Boston, MA; Richard Rhodes, B.S., M.S., Advance Pressure Garment Technology Development Lab, Crew Survival and Space Suit Systems Branch EC5, NASA Johnson Space Center, Houston, TX; and Leia Stirling, M.S., Ph.D., Aeronautics and Astronautics, Institute for Medical Engineering and Science, Massachusetts Institute of Technology, Cambridge, MA.

REFERENCES

1. Abe D, Yanagawa K, Niihata S. Effects of load carriage, load position, and walking speed on energy cost of walking. Appl Ergon. 2004; 35(4):329–335. [Accessed 2016 March 22]. Available from: http://www.sciencedirect.com/science/article/pii/S0003687004000572.
2. Abramov I, Moiseyev N, Stoklitsky A. Concept of space suit enclosure for planetary exploration. SAE Conference [Internet]. 2001 [Accessed 2014 March 18]. Available from: http://spacecraft.ssl.umd.edu/design_lib/ICES01-2168.suit_enclosure.pdf.
3. Atwell B, Boeder P, Ross A. Space radiation analysis for the Mark III spacesuit. 2013 [Accessed 2014 March 18]. Available from: http://arc.aiaa.org/doi/pdf/10.2514/6.2013-3381.
4. Cowley MS, Margerum S, Hharvill L, Rajulu S. Model for predicting the performance of planetary suit hip bearing designs. 2012 [Accessed 2016 May 4]. Available from: http://ntrs.nasa.gov/search.jsp?R=20130000596.
5. Cullinane C, Rhodes R, Stirling L. Mark III technology demonstrator suit: hip brief assembly bearing analysis. In: International Conference on Environmental Systems; 2015; Bellevue, WA. Stafford Springs (CT): International Conference on Environmental Systems, Inc.; 2017.
6. Gast MA, Moore SK. A glimpse from the inside of a space suit: what is it really like to train for an EVA? Acta Astronaut. 2011; 68(1-2):316–325. [Accessed 2016 Feb. 29]. Available from: http://www.sciencedirect.com/science/article/pii/S0094576510003036.
7. Gernhardt M, Jones J, Scheuring R, Abercromby F, Tuxhorn J, Norcross J. Risk of compromised EVA performance and crew health due to inadequate EVA suit systems. In: McMurphee J, Charles J, editors. Human Health and Performance Risks of Space Exploration Missions. Houston (TX): NASA Lyndon B. Johnson Space Center; 2009. NASA SSP-2009-3405.
8. Grasso R, Bianchi L, Lacquaniti F. Motor patterns for human gait: backward versus forward locomotion. J Neurophysiol. 1998; 80(4):1868–1885. [Accessed 2016 March 21]. Available from: http://jn.physiology.org/content/80/4/1868.short.
9. Kosmo J, Ross A. Space suit mobility evaluations in lunar/Mars gravity environments. Warrendale (PA): SAE; 1998. SAE Technical Paper 981627.
10. LaFiandra M, Wagenaar RC, Holt KG, Obusek JP. How do load carriage and walking speed influence trunk coordination and stride parameters? J Biomech. 2003; 36(1):87–95. [Accessed 2016 March 22]. Available from: http://www.sciencedirect.com/science/article/pii/S0021929002002439.
11. Mitchell K, Ross A, Blanco R, Wood A. Hazard analysis for the Mark III Space Suit Assembly (SSA) used in one-g operations. 2012 [Accessed 2014 March 18]. Available from: http://ntrs.nasa.gov/search.jsp?R=20120017944.
12. Nelson AJ. The functional ambulation performance of elderly fallers and non-fallers walking at their preferred velocity. NeuroRehabilitation. 1999; 13(3):141–146.
13. Newman D, Barratt M. Life support and performance issues for extravehicular activity (EVA). In: Churchill S, editor. Fundamentals of Space Life Sciences. Melbourne (FL): Krieger Publishing Co.; 1997.
14. Opperman RA, Waldie JM, Natapoff A, Newman DJ, Jones JA. Probability of spacesuit-induced fingernail trauma is associated with hand circumference. Aviat Space Environ Med. 2010; 81(10):907–13. [Accessed 2014 March 31]. Available from: http://openurl.ingenta.com/content/xref?genre=article&issn=0095-6562&volume=81&issue=10&spage=907.
15. Powers CM. The influence of abnormal hip mechanics on knee injury: a biomechanical perspective. J Orthop Sports Phys Ther. 2010; 40(2):42–51.
16. Ross A, Rhodes R, Dover ILC. Z-2 prototype space suit development. 44th International Conference on Environmental Systems; July 13–17, 2014; Tuscon, AZ. Stafford Springs (CT): International Conference on Environmental Systems, Inc.; 2014:1–11.
17. Scheuring RA, Mathers CH, Jones JA, Wear ML. Musculoskeletal injuries and minor trauma in space: incidence and injury mechanisms in U.S. astronauts. Aviat Space Environ Med. 2009; 80(2):117–124. [Acessed 2014 March 31]. Available from: http://openurl.ingenta.com/content/xref?genre=article&issn=0095-6562&volume=80&issue=2&spage=117.
18. SolidWorks Corporation. SolidWorks. Waltham (MA): Dassault Systèmes; 2016.
19. Sutherland D. The development of mature gait. Gait Posture. 1997; 6(2):163–170. [Accessed 2016 March 9]. Available from: http://www.sciencedirect.com/science/article/pii/S0966636297000295.
20. Thorstensson A. How is the normal locomotor program modified to produce backward walking? Exp Brain Res. 1986; 61(3):664–668. [Accessed 2016 March 21]. Available from: http://link.springer.com/10.1007/BF00237595.
21. Wade M. NASA Mark III [Internet]. Encyclopedia Astronautica. [Accessed 2015 Dec. 13]. Available from: http://www.astronautix.com/n/nasamarkiii.html.
22. Wehner M, Quinlivan B, Aubin PM, Martinez-Villalpando E, Baumann M, et al. A lightweight soft exosuit for gait assistance. In: IEEE International Conference on Robotics and Automation (ICRA); May 6−10, 2013; Karlsruhe, Germany. New York: IEEE; 2013:3362–3369.
23. Whittle MW, Levine D. Three-dimensional relationships between the movements of the pelvis and lumbar spine during normal gait. Hum Mov Sci. 1999; 18(5):681–692. [Accessed 2016 March 7]. Available from: http://www.sciencedirect.com/science/article/pii/S0167945799000329.