Seismic: Biomechanical Engineering Team
University of Twente

Masters Internship

Quantifying Movement Quality and Stability in Seismic Tech

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Introduction

Seismic is a California-based company specializing in powered clothing. Their products are a marriage of form-fitting textiles, electromechanical motors, and embedded sensors designed to provide targeted assistive forces during movements or stationary tasks. Though they are similar in some aspects to traditional exoskeletons, they differ in two fundamental ways. First, Seismic products lack rigid links, drastically reducing weight and allowing for the products to be worn under normal clothing. Secondly, while most traditional exoskeletons work by providing assistive torques at the joints of interest, powered clothing provides assistive forces across joints in a manner mirroring the natural function of muscle-tendon units.

Currently, Seismic's primary product is a core wellness suit that provides support at the hips and lower back (Figure 1). It is targeted towards active, older adults who desire support in their daily lives. A suite of rehabilitation garments for boys with Duchenne muscular dystrophy (DMD) is also currently in development. The three products in this line aim to provide therapeutic stretching at the ankle, trunk and hip support, and arm support during reaching tasks.

Figure 1: Seismic's core wellness suit (Photo credit: Wes Sumner)

Given the products' intended use, it is important to be able to quantify users' movement quality and stability both with and without support from a powered garment. Doing so enables the Seismic team to evaluate a product's efficacy with quantifiable and easy-to-compare metrics, to target and scale assistive forces based on biomechanical user data, to understand more completely the effects of their products on body kinematics, and to tailor their development process based on these findings.

Problem Specification

The primary challenge with quantifying movement quality at Seismic is the lack of available sensors with which to collect movement data. 9-axes inertial motion units (IMUs) are the main motion capture sensor used on-board Seismic's garments. Three are used on the core wellness suit– one located on the lumbar spine and two located laterally on each thigh (Figure 1). On the DMD products, a single IMU is placed on each body segment of interest. The ankle stretch device, for instance, has an IMU placed anteriorly on the shank and another under the foot at the second metatarsal head. Useful information may also be obtained from the suits' force and current sensors which track the output of the motors. Additionally, tools such as timers, cameras, goniometers, pressure sensors, and other simple devices and instruments commonly used in physical therapy settings, hardware labs, and fashion studios are available on-site at Seismic's headquarters.

The diverse target users also pose a unique challenge. Many of these populations would be considered risk-sensitive in traditional research environments. Assessments performed on these groups must therefore account for potentially increased risk due to age-related changes in the neuromuscular system, diminished muscular endurance and force output, lower stability, or decreased range of motion.

Additionally, there are questions users want answered regarding their powered clothing use. They want to know how using the core wellness suit increases their endurance so they can golf longer or how it helps them rise from a chair; or they want to know how the DMD products help to maintain and improve ankle range of motion or how they support arm function so users can feed themselves.

Based on these and internal questions regarding suit efficacy, biomechanical pilot studies were done that examine the following:

- Balance
- Endurance
- Trunk stability
- Hip function
- Reach quality
- Walking quality

Pilot Studies

The pilot studies discussed below were all performed at Seismic's headquarters in Menlo Park, California. The actions performed were either already part of the user testing team's standard test protocol or easily integrated into it. For each study, the protocol can be found in the appendix. All of the pilot studies utilized a single, trunk-mounted IMU tethered to a laptop for data collection. Data analysis was performed using custom Python code[1].

Additionally, many of the studies and metrics produced correspond well to assessments commonly used in physical therapy, allowing for comparisons to be made between individual user data and pre-existing

[1] Because the folks at Mathworks are extortionists.

normative datasets. This also allows for user data to be mapped to performance levels[2] commonly used within the rehabilitation community, even if the primary users are healthy, active individuals.

The pilot studies completed include:

- Functional Reach Test
- Sway Test
- 10-Meter Walk Test

Functional Reach Test

The Functional Reach Test (FRT) is commonly used in physical therapy to assess balance and functional motion. In this test, the distance a subject can reach forward beyond arm's length without taking a step is recorded [1]. The test is simple to perform and easy to adapt for users of various ability, making it ideal for many diverse user populations. Additionally, numerous studies have shown the reliability and validity of this test in identifying balance deficits and predicting fall risk in older adults [1]. Two important metrics can be derived from the FRT: the maximum distance reached and the smoothness of the movement.

Implementation

At Seismic, the FRT was used primarily to assess trunk stability in boys with DMD. The procedure for the modified FRT, in which the subject performs the test while seated, was thus also included as an addendum to the protocol for subjects who are not capable of standing for extended periods.

The test was also conducted with the addition of an IMU attached to the upper thoracic spine with an elastic strap. In doing so, more information on the stability of the trunk throughout the movement and on how the user moves may be obtained. To my knowledge, this is the first time such an assessment has been performed within the DMD population. Previous studies looked only at trunk displacement while performing select items of the Performance of Upper Limb (PUL) Test [2].

Distance

The primary measurement outcome of the FRT is maximum distance achieved, which is a powerful predictor of fall risk. In older adults, a forward reach of 25.40 cm (10 in) or greater is considered normal. Adults with a reach between 15.24 cm - 17.78 cm (6 in - 7 in) are considered to have limited functional balance, and their fall risk is two times greater than other adults who have not experienced a fall in the past. This risk continues to grow as the maximum distance decreases. With a reach of less than 15.24 cm (6 in), the associated fall risk is four times greater [1].

[2]Examples include: the categories "non-fall-prone" and "fall-prone" commonly seen in stability studies and the "community ambulator", "household ambulator", and "physiological ambulator" categories based on walking speed used in stroke rehabilitation.

Less information on norms for the FRT exist for pediatric populations, and most of the studies conducted on non-typically developing children involve children with cerebral palsy. For typically developing (TD) children, performance on the FRT varies up until age 11, when max reach values stabilize [3]. Average scores for these children are shown in Table 1. Variance in the scores is primarily affected by age, which itself accounts for much of the variance due to height and weight [3].

Age (yr)	*Mean (cm)*	*95% CI (cm)*	*Critical Reach (cm)*
5-6	21.17	16.79-24.91	16.79
7-8	24.21	20.56-27.96	20.57
9-10	27.97	25.56-31.64	25.56
11-12	32.79	29.68-36.18	29.68
13-15	32.30	29.58-36.08	29.58

Table 1: Mean values, confidence interval, and critical reach in the pediatric FRT by age group. Scores below the critical reach value indicate increased fall risk [3].

In children with DMD, performance on the FRT has been correlated with functional level on the Brooke Scale for the lower extremity in at least one study [4]. The study showed no significant difference in scores on the modified FRT between children of a functional level up to Level III and TD peers. However, in the standard FRT, children of a functional level of III performed significantly worse than TD peers, with an average score of 15.20 cm (5.98 in) compared to 22.35 cm (8.79 in) in TD children.

Movement Smoothness

Smoothness is a metric that can be used to quantify the quality of many movements. For non-cyclic motions, smoothness is "related to the continuality or non-intermittency of a movement" [5]. In movements, intermittency is defined as alternating between periods of deceleration and acceleration and is evidenced by multiple peaks in the speed profile. For smooth, non-cyclic motions the speed profile is typically bell-shaped with a single peak. For cyclic movements such as walking, overall smoothness is a function of the smoothness of each individual, distinct component of the motion [5].

Several measures exist to quantify movement smoothness, with the most common ones being jerk-based measurements. Another common method is to count the number of local maxima within a speed profile [5].

During the FRT pilot tests, the smoothness of the trunk movement[3] was recorded using an IMU. The anterior-posterior (A-P)[4] component of the acceleration data[5] was compared to a reference dataset from a healthy, internal user to determine the key phases of the reach (Figure 2). For a forward reach, periods

[3]Not the smoothness of the reach, which would require the IMU to be placed on the hand instead.

[4]For this and subsequent pilot tests, only the A-P components of the data are shown and discussed. Both the medio-lateral (M-L) and longitudinal components were also studied but are omitted here for brevity. In general, it is important to analyze all three components of a movement as each provides a slightly different perspective on the quality of the movement.

[5]The movement data in these pilot studies were filtered using a 2nd-order lowpass Butterworth filter with a cutoff frequency of 6 Hz to eliminate any artifacts from environmental noise sources such as the motors.

of movement are indicated in the acceleration profile as sloped lines[6] (Figure 2 sections B and D) while periods of inaction are indicated by relatively flat lines (Figure 2 sections A and C).

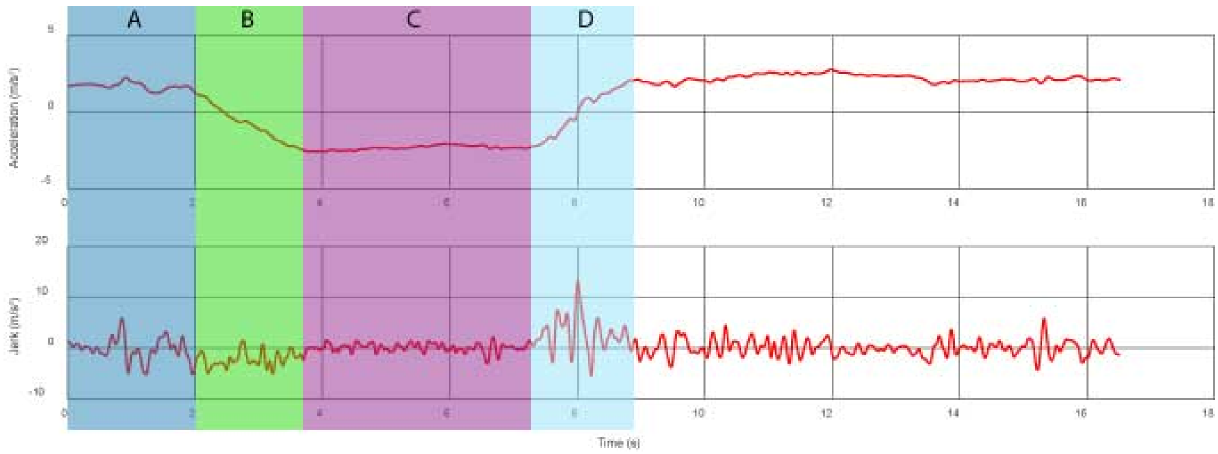

Figure 2: The A-P component of the acceleration (top plot) and jerk (bottom plot) during the forward FRT for a boy with DMD. The shaded areas correspond to the different reach phases: A) quiet standing, B) reach, C) hold, and D) return to standing. Note, this particular user raises his arms and reaches in one movement instead of raising his arms, pausing, and then reaching. Also, standing acceleration values are not 0 because the user stands with his upper back curved backwards and the gravity component has not been removed.

Looking at the jerk provides deeper insight into the quality of the reaching movement in each phase. Increased jerk is a hallmark of decreased movement smoothness, and, when it occurs in particular phases, may indicate portions of the movement which are more difficult for the user to perform. For instance, in Figure 2, the magnitude of the jerk is much higher during the return phase than in the preceding ones, which may be because the user finds it more difficult to initiate movement from an extended position. There is also increased jerk at the end of the standing phase due to the compensatory movement needed to raise the arms and begin the reach.

Sway Test

Postural sway, defined as the horizontal movement of the body's center of gravity (COG), has often been correlated with stability. Even during quiet standing in healthy subjects, sway occurs due to changes in the location of the body's COG from breathing and the muscles' inability to produce a steady, constant force due to neuromuscular limitations. To maintain balance, the body's COG must remain within the base of support (BOS), so, to this end, postural sway should be minimized. When examining sway, it is important to consider the velocity of the body center of pressure (COP) and the range of the motion.

[6]Which makes sense, as the derivative of a bell (parabola) is a line.

Implementation

In research settings, postural sway is usually measured with sway tests. During a sway test, the subject stands quietly while movement data is collected. The instrument typically used during these tests is a force plate. An IMU may also be used though, as was the case here.

This test was used at Seismic with older adults and boys with DMD, as both populations have a keen interest in improving their balance. Measurements were taken both with the core wellness suit[7] off and with it engaged. Here, increased postural stability is evidence of the efficacy of the standing support function, and, for users, may translate into improved balance and increased endurance during static and dynamic standing activities such as waiting in lines or performing assembly line tasks.

Notably, the ability to stand for extended periods may be diminished in both of the test populations. For this reason, the duration of the test was limited to 30 seconds[8]. Doing so limits fatigue, decreases the effects of waning user motivation[9] or outside audiovisual stimuli[10], and lowers the risk to users. Modified versions of the sway test were also created for users with DMD who are unable to stand. These include a version where the boy sits up in his wheelchair, unsupported by the chair's back, and a more difficult one where he sits on a yoga ball with his feet flat on the ground. While the original sway test provides information on balance methods employed both at the hip and at the ankles to control the position of the total body's COG, these methods only give insight into mechanisms of postural control employed at the hip to control the COG of the HAT[11] segment so should only be used when deemed necessary for safety.

The standard protocol also includes options for standing with eyes closed, standing on a balance pad, and standing with eyes closed while on a balance pad. These represent three increasingly difficult standing tasks– standing without visual feedback, standing without proprioceptive feedback, and standing with neither, respectively– and were conducted to gain deeper insight into the efficacy of Seismic technologies' standing support function. In healthy users, changes in sway measures between the suit-off and the suit-engaged trials may be too small to attribute to the standing support function rather than normal variation. I hypothesized that a greater change between conditions–one too large to be random– may occur when the subject's balance is diminished. More people must be tested though[12] to prove this hypothesis. Moreover, testing these additional conditions allows for performance levels to be created for users who desire a non-binary rating for their balance[13].

Stabilograms

A common method of quantifying sway is to track the movement of the body's COP during quiet standing with a force plate measuring the ground reaction force (GRF). By doing, so the coordinates of the COP

[7] Or its DMD counterpart depending on the user.
[8] This is also the shortest duration found in a quick literature review.
[9] Some people, at longer testing times, got bored and began talking or humming.
[10] For instance, people passing in the hallway.
[11] Head-Arms-Trunk
[12] The core wellness suit is a bespoke product so there were very few healthy, young subjects who had a suit that could be tested.
[13] Currently, sway tests most often group people into categories of either not-at-risk or at-risk (of falling). Additional testing may lead to the creation of additional "good", "better", or "best" type levels within the not-at-risk group.

can be calculated using the equations

$$x = \frac{-hF_x - M_y}{F_z} \tag{1}$$

$$y = \frac{-hF_y - M_x}{F_z} \tag{2}$$

where h is the height of any material that may be covering the plate; F_x, F_y, and F_z are the three components of the GRF; and M_x and M_y are the x- and y-components of the ground reaction moment, respectively. The coordinates can then be plotted to produce a stabilogram, an example of which is seen in Figure 3.

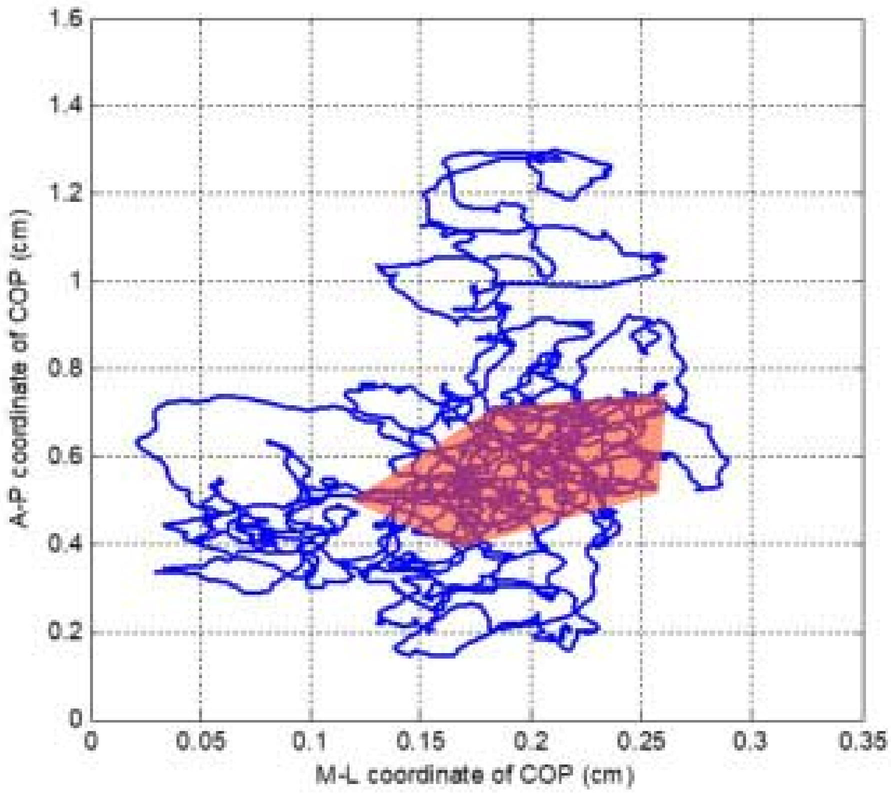

Figure 3: An example stabilogram taken from a 90-sec trial of a person standing quietly. The x-axis shows the medio-lateral movement of the COP, while the y-axis shows the anterior-posterior movement. The red polygon highlights a focal area within the data. Note, the path has been translated into quadrant I so that multiple plots can easily be compared [6].

Changes in sway–and by extension, stability–can easily be seen in stabilograms and are evidenced by increases or decreases in the area of the plot. Better stability is associated with a smaller plot area, while decreased stability is associated with a larger plot area. Focal areas are often evident in stabilograms, as seen in the highlighted region in Figure 3. These focal areas correspond to regions the COP tends to settle into, and, as before, a smaller area is associated with better balance. The axis along which a change occurs can also provide useful information on the strategy employed to maintain balance. Changes along the anterior-posterior (A-P) axis typically correspond to an ankle-based balance strategy. In this strategy, joint torques are varied about the ankles to maintain the COG's position within the BOS. An ankle-based strategy is usually used for small postural adjustments. Changes along the medio-lateral (M-L) axis correspond to a hip-based strategy, in which joint torques are varied about the hip to adjust the position of the COG [7]. This type of strategy is used when shifting weight from one leg to the other or in response to a large, external perturbation.

Another useful quantity that can be calculated from the COP coordinates is the average velocity of the COP. Generally, a larger average velocity indicates increased movement of the COP and is associated with worse balance and stability. Conversely, a lower average velocity is associated with less movement and better balance and stability.

Sway Diagrams

Because stabilograms can only be produced with the use of a force plate, which are often bulky, expensive, and difficult to calibrate, they may be an inaccessible or impractical measure outside of a biomechanics lab. Additionally, this reliance on force plates means they can not be produced "in the wild", such as at a user's An alternative in this case would be to relate the body's sway to its acceleration instead of the movement of the COP. By doing so, a single, body-mounted tri-axial accelerometer can be used in place of a force plate. Unlike force plates, accelerometers are small, cheap, and virtually ubiquitous, being found in smartphones and many other personal electronic devices. This makes them better suited for clinical or in-home measurements.

When using an accelerometer to quantify postural sway, the body is modeled as a simple, inverted pendulum, pinned at the ankles. The accelerometer, placed above the hips on the trunk, is then essentially an inclinometer capturing the body's tilt. Additionally, the model assumes the body's sway is small so that the height of the accelerometer from the ground remains constant.

In this method, the variable of interest is the extension of the acceleration vector to the ground, $\bar{D}$. This is found by first calculating the magnitude of the resultant acceleration, $|\bar{A}|$, from the 3 measured components of the acceleration as shown in (3). The angle each component makes with the resultant– α, β, and γ for x, y, and z respectively (Figure 4)– is then calculated from the directional cosine ((4), (5), (6)).

$$|\bar{A}| = \sqrt{{a_x}^2 + {a_y}^2 + {a_z}^2} \tag{3}$$

$$\alpha = \arccos\left(\frac{a_x}{|\bar{A}|}\right) \tag{4}$$

$$\beta = \arccos\left(\frac{a_y}{|\bar{A}|}\right) \tag{5}$$

$$\gamma = \arccos\left(\frac{a_z}{|\bar{A}|}\right) \tag{6}$$

Because the postural sway is assumed to be small, the z-components of both $\bar{A}$ and $\bar{D}$ are equivalent (Figure 4) and the angles they form with their respective resultants are also equal, namely γ. Knowing this and the height of the accelerometer from the ground, d_z, the magnitude of $|\bar{D}|$ can be calculated using (7). The x- and y-components of $\bar{D}$ can then be calculated with (8) and (9).

$$|\bar{D}| = \frac{d_z}{-\cos\gamma} \tag{7}$$

$$d_x = |\bar{D}|\cos\alpha \tag{8}$$

$$d_y = |\bar{D}|\cos\beta \tag{9}$$

Note, $\bar{D}$ and its components are given in units of length, not acceleration. Though $\bar{A}$ and $\bar{D}$ are collinear, $\bar{D}$ may be better understood as a vector describing the orientation of $\bar{A}$ as extended from the IMU housing to the ground. The z-component of $\bar{D}$ has a magnitude equal to the height of the accelerometer off of the ground and is collinear to a_z. $\bar{D}$'s x- and y-components have magnitudes equal to the distance of $\bar{D}$'s head from the BOS in the horizontal plane and are parallel to a_x and a_y, respectively. Their displacement, known as the sway path, is a description of how the body's sway changes throughout the testing period.

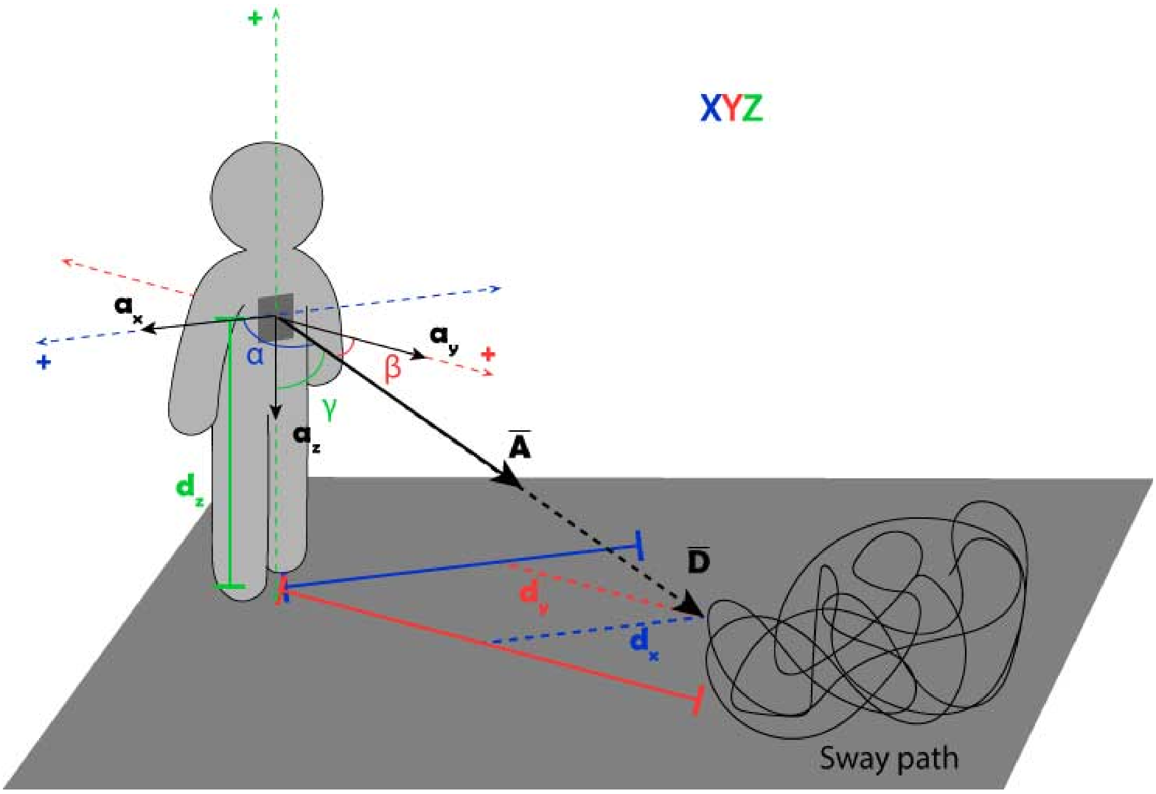

Figure 4: Diagram showing the resultant acceleration, $\bar{A}$; its constituent components, a_x, a_y, and a_z; the extension of the resultant, $\bar{D}$; its components, d_x, d_y, and d_z; and the angles between the components and their resultants, α, β, and γ. The sway path of $\bar{D}$ is produced by plotting the horizontal components of $\bar{D}$, d_x and d_y, as they vary with time [6].

This sway path may be visualized by plotting d_x versus d_y. Doing so produces a plot similar to a stabilogram (Figure 5). This plot, a sway diagram, is interpreted in a similar manner to a stabilogram, with changes in plot area associated with increased or decreased balance and stability. Like the stabilogram, the average velocity of the sway can also be calculated. As before, higher velocities indicate lower balance and stability and lower velocities indicate higher balance and stability.

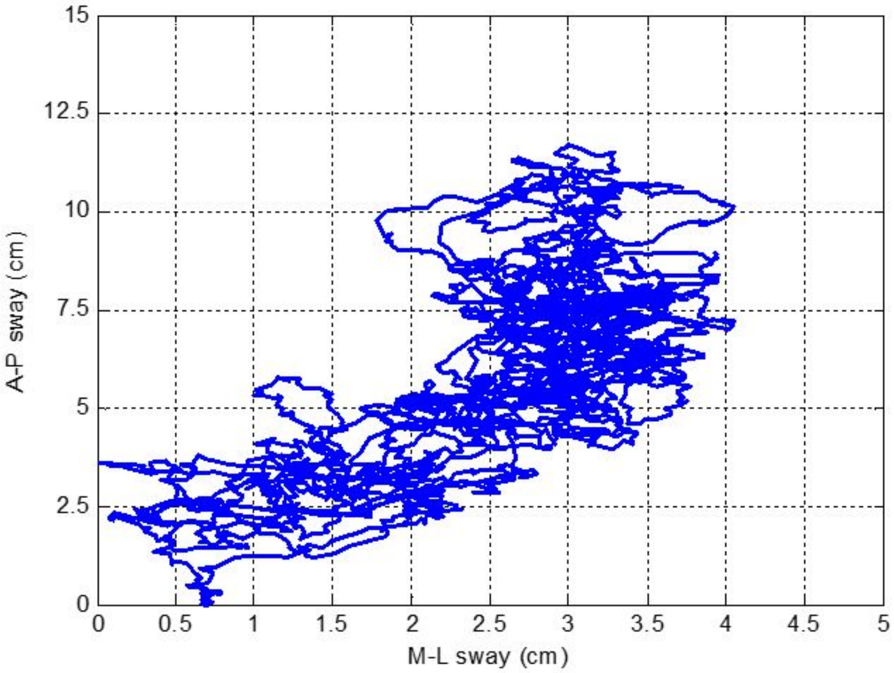

Figure 5: An example sway diagram taken from a 90-sec trial of a person standing quietly. The x-axis shows the medio-lateral movement of $\bar{D}$, while the y-axis shows the anterior-posterior movement [6]

It should be noted that, though the numbers and diagrams produced using this method are analogous to those obtained using the COP and a force plate, they will not be the same. Both methods quantify body sway, but they do so by making different assumptions and by measuring two different variables– the movement of the COP for stabilograms and the movement of a vector collinear to the resultant acceleration for sway diagrams– at different locations. Initial studies have found though that, when using a force plate and accelerometer to distinguish between different standing conditions, both methods are capable of discerning different conditions and produce results that mirror each other [6] [8], as seen in Table 2.

$\bar{v}_{ave}$ *(cm/s)*	*Eyes open*	*Eyes closed*	*Eyes open w/ pad*	*Eyes closed w/ pad*
COP	0.80	0.72	27.04	48.32
Sway	10.80	10.52	12.66	14.21

Table 2: Mean velocities for the COP and sway under different standing conditions [6]

Meaningful Numbers

In older adults and populations with compromised balance, sway measurements are often used as predictors of fall risk. Previous research has shown that body sway increases with age and that the likelihood of falling increases with increasing sway. Additionally, mean COP velocity tends to be higher in older adults who have experienced a fall than those who have not [9].

In relation to Seismic technology, particular attention should be paid to changes in the M-L direction. Because Seismic's current line of products provide support at the hips and lower back and do not cross the ankles, greater changes may be seen in this direction than the A-P since changes in the M-L direction are associated with hip-based balance strategies, as previously stated.

Additionally, categorizing individual balance into scoring brackets such as "good", "ok", and "bad" may be of particular interest for Seismic users, many of whom belong to fall-prone populations. To do so, a reference data set which individual data can be compared to must be established. For older adults, a lot of data exists, especially with regards to falling and assessing fall risks. For boys with DMD, no such data set currently exists and, due to limited resources, it is not possible to create this data set within Seismic. Instead, a general comparison may be made to healthy, young subjects performing tasks of varying difficulty. For instance, Jebelli et al., in a study involving construction workers wearing a full harness [10], found an average sway velocity of about 5 mm/s (0.197 in/s) when standing compared to about 18 mm/s (0.709 in/s) when squatting[14]. The same studies reported sway velocities of about 22 mm/s (0.866 in/s) and 38 mm/s (1.50 in/s) for standing and squatting, respectively, while carrying a load weighing up to 8.16 kg (18 lbs)[15].

10-Meter Walk Test

Walking, as the primary form of human locomotion, is important to examine, and many tests and metrics exist within the clinical space and academia that look at walking to quantify fall risk, functional mobility, dynamic stability, and movement quality. One such commonly used test is the 10 Meter Walk Test (10 MWT), in which a subject walks 10 m (32.8 ft) across a level surface while timed. Like the FRT, the 10 MWT is easily implemented, cheap, and versatile, as it can be done at either the subject's preferred walking speed or as quickly as safely possible and even with an assistive device. Additionally, as a standard assessment performed within physical therapy, norms and data for the test exist for many populations.

When using the 10 MWT to quantify movement quality, some of the meaningful quantities that can be obtained include the average walking speed, spatial and temporal step symmetry, phasing, and variability.

Implementation

The 10 MWT was used primarily for boys with DMD. Like the FRT, an IMU mounted to the upper thoracic spine was included to provide additional insight into the movement. Video was also taken so

[14] Here, squatting is considered to require more balance than standing.
[15] Here, this is considered a high fall-risk task compared to the one discussed previously.

that step count[16] could be quantified and missteps or gait abnormalities could be visually identified post-test.

Additional gait parameters beyond the walking speed are typically derived from motion capture data or, at the very least, from two IMUs placed on the dorsal aspects of the feet. At Seismic, however, no motion capture system is available and the IMU software is only capable of collecting from a single IMU at once. These limitations lead to the development of a novel method for identifying gait events (see Appendix). To validate this method and to create a dataset that individual user data could be compared to, 10 MWT data was collected from internal subjects as well. A longer walk with a single user was also conducted to test the robustness of the sensor placement method and its effects on data quality (particularly in the M-L direction). In this walk, performed with a subject who had no known history of leg injuries or surgeries, steps over a relatively level surface were counted and then compared to the number identified via the novel method. A normal arm swing was not used, however, as the subject was tethered to a laptop and this needed to be carried. Both arms were used to carry the laptop to maintain bilateral symmetry. Turning was also limited as this is known to produce differences in gait variables in the right vs the left legs. From these preliminary tests, the method appears to be viable at least for non-pathological gait.

Average Walking Speed

The average walking speed during the 10 MWT is computed with the following equation:

$$v_{ave} = \frac{10}{\Delta t} \tag{10}$$

where 10 is the distance covered in meters and Δt is the total time taken to cover that distance. For quantifying movement quality and the efficacy of Seismic technology, it is important to look at both the magnitude of this number and changes in the magnitude between trials of different conditions, i.e. suit inactive vs suit active.

For healthy older adults, the average walking speed, v_{ave}, and the change in the average walking speed can be compared to norms that exist either for adult populations (18 - 64 years of age) or for elderly adult populations (+64 years old), depending on the age of the subject. Normative v_{ave} values for men ages 20 - 99 years old are shown in Table 3. Values for women are shown in Table 4 [11]. For elderly adults, a change in velocity of 0.05 m/s (0.164 ft/s) is considered small but meaningful. A change of 0.10 m/s (0.328 ft/s) is considered substantial. In terms of the minimally clinically important difference, these values are 0.05 m/s (0.164 ft/s) for a small change and 0.13 m/s (0.427 ft/s) for a significant one [12].

These values may further be correlated with levels of functional walking ability, which are often referred to in stroke rehabilitation literature. Broadly, these levels are "community ambulator", "household ambulator", and "physiological ambulator", with independence being the primary differentiator between them. Table 5 below shows the walking speeds associated with each level. At the low end of the highest functional level, community ambulators can independently enter and exit their homes and can negotiate curbs but may require assistance in stores or shopping centers [13]. It is associated with an average walking speed of at least 0.4 m/s (1.312 ft/s). Household ambulators are those who walk at speeds less

[16] And any derived quantities like cadence

Age (decade)	*Sources (n)*	*Subjects (n)*	v_{ave} *(m/s)*	*Grand mean (95% CI) range*
20s	10	155	1.358	1.217 - 1.474
30s	5	83	1.433	1.320 - 1.538
40s	4	96	1.434	1.270 - 1.470
50s	6	436	1.433	1.122 - 1.491
60s	12	941	1.339	1.033 - 1.590
70s	18	3671	1.262	0.957 - 1.418
80s	10	1091	0.968	0.608 - 1.221

Table 3: Mean values and ranges for the 10 MWT for men given by age decade. Number of source articles and total subjects per decade are also given [11].

Age (decade)	*Sources (n)*	*Subjects (n)*	v_{ave} *(m/s)*	*Grand mean (95% CI) range*
20s	11	1091	1.341	1.082 - 1.499
30s	5	180	1.337	1.256 - 1.415
40s	7	104	1.390	1.220 - 1.420
50s	10	456	1.313	1.100 - 1.555
60s	17	5013	1.241	0.970 - 1.450
70s	29	8591	1.132	0.830 - 1.500
80s	17	2151	0.943	0.557 - 1.170

Table 4: Mean values and ranges for the 10 MWT for women given by age decade. Number of source articles and total subjects per decade are also given [11].

than 0.4 m/s (1.312 ft/s) [14] but still rely to some extent on walking for household activities. At the lowest functional level, physiological walkers are largely dependent on wheelchairs for mobility and may only walk for exercise, either at home or during physical therapy with parallel bars [13]. No normative speed data exists for this group as they are unable to perform the test without assistance from another person.

The highest two levels can be further divided into sub-levels. At the highest functional level, "full community ambulators" are those capable of average walking speeds greater than 1.2 m/s (3.937 ft/s). They are independent within their homes and at moderate community activities and can safely cross streets and negotiate uneven terrain without assistance. "Least-limited community ambulators" are also independent at moderate community activities, but may require some assistance in crowded shopping centers or need supervision in local stores or uncrowded shopping centers [13]. They have an average walking speed between 0.8 m/s to 1.2 m/s (2.625 ft/s to 3.937 ft/s). "Most-limited community ambulators" have an average walking speed between 0.4 m/s to 0.8 m/s (1.312 ft/s to 2.625 ft/s) [14]. Household ambulators are further divided into either "unlimited" or "limited". Limited househould ambulators can perform all household activities without a wheelchair but may require supervision within the bedroom or bathroom or when leaving the home. They experience some difficulty with stairs, curbs, and uneven terrain.

Functional Level	*Walking Speed (m/s)*
Full community ambulator	> 1.2
Least-limited community ambulator	0.8 - 1.2
Most-limited community ambulator	0.4 - 0.8
Household ambulator	< 0.4
Physiological ambulator	—-

Table 5: Functional walking levels and associated walking speed ranges. A range for physiological ambulators is not included because they are unable to perform the 10 MWT without assistance [13] [14].

Walking Symmetry

Symmetry is another metric that may be used in conjunction with velocity to quantify gait quality. Unlike velocity, changes in gait symmetry are not sensitive to age so gait symmetry is a robust measure across different age groups [15]. To examine gait symmetry, however, a reliable method of discerning steps and the key phases and events of the gait cycle is needed (see appendix). The method employed should be able to distinguish between stance phase and swing phase, to identify periods of single support or double support, and to quantify step length and step width[17].

The general formula used to determine the symmetry index, SI, is:

$$SI = \frac{X_1 - X_2}{X_1 + X_2} \tag{11}$$

where X_1 is the parameter of interest (i.e. double support time or step length) for the dominant leg[18] and X_2 is for the contralateral leg [16]. Values of 0 indicate perfect symmetry. Positive index values indicate that the magnitude of the examined parameter is greater for the dominant leg than it is for the contralateral leg, whereas negative values indicate the reverse.

When examining symmetry, it is important to look at both temporal and spatial symmetry. However, at Seismic there are currently no means to examine the spatial parameters of gait, such as step length or center of oscillation[19]. Regarding temporal symmetry, parameters typically examined are swing time and stance time[20].

[17] Alternately, the spatio-temporal properties of stride can be examined, but to look at both step and stride would be redundant.

[18] It is left as an exercise to the reader to determine how to define leg dominance as this often varies depending on the task being examined (i.e. a stability task vs a skilled task) or the conventions of the field. Stroke researchers, for instance, often define the dominant leg as the non-hemiparetic one. Studies with split-belt treadmills, however, define the dominant leg as the one placed on the faster belt.

[19] Which is derived from step length.

[20] Here, it is possible to look at the entire duration of stance, the single support time, or the double support time. Again, when just examining symmetry, looking at more than one of these provides redundant information. The author, though, prefers to and recommends that Seismic should look at double support time. This is because evidence of motor learning

It is also possible with a force plate[21] to evaluate symmetry in terms of the average force profile of a right versus a left step. When doing so, the magnitude and duration of the loading response, the magnitudes of the braking and propulsive forces, and the magnitude of the impact transient can be compared using Equation 11.

Ideal gait exhibits symmetry indexes close to zero, with interlimb differences of up to 6% in temporal parameters and GRFs still considered to be within the normal range. For spatial symmetry, the boundary between healthy and pathological gait is not as clear. A 10% difference is a common, although arbitrary, cut-off value used in research. Patterson et al. [18] report a cut-off value for step length symmetry of 1.08[22]. The same work reports 1.06 and 1.05 as cut-off values for swing time symmetry and stance time symmetry, respectively [18].

Beyond identifying the severity of asymmetry and the gait parameters which exhibit gait asymmetry, further interpretation of the results for a single subject is not necessarily straightforward, especially without knowledge of the person's medical history. Gait asymmetry may be a sign of insufficient neuromuscular control, weakness or injury in one leg, or the use of an inefficient gait pattern. It is important to recognize that there may be many possible causes for an asymmetry. For instance, greater braking forces on the right leg could indicate that the left is accelerating the body more during swing so greater forces are needed to slow the COM or that the person has diminished balance during stance on that (the left) leg.

Phasing

Phasing describes the phase shift in limb displacement during walking. It is usually obtained from the cross-correlation of displacement signals[23] taken from both the left and right legs [19]. During normal symmetric walking, a shift of half of a period is expected since the legs move reciprocally Figure 6. A shift of any other value is indicative of a temporal asymmetry. At a phase shift of a full period, both legs are moving at the same time.

and adaptation are seen in interlimb parameters. Intralimb parameters like single support time and stride length change quickly in response to new motor patterns but exhibit no evidence of learning or retention upon subsequent exposure to the novel pattern [17]. Because gait patterns are highly specific and Seismic's products are meant to be worn repeatedly for long periods of time, how users learn and adapt new motor patterns in response to garment use may be of interest to Seismic in the future.

[21] Which the author hopes Seismic will obtain soon.

[22] Here, the symmetry index is calculated as a simple ratio between the right and the left, with the larger number placed in the numerator. Equation 11, however, returns normalized values.

[23] These signals can be position-related value such as ankle position, limb angle, etc.

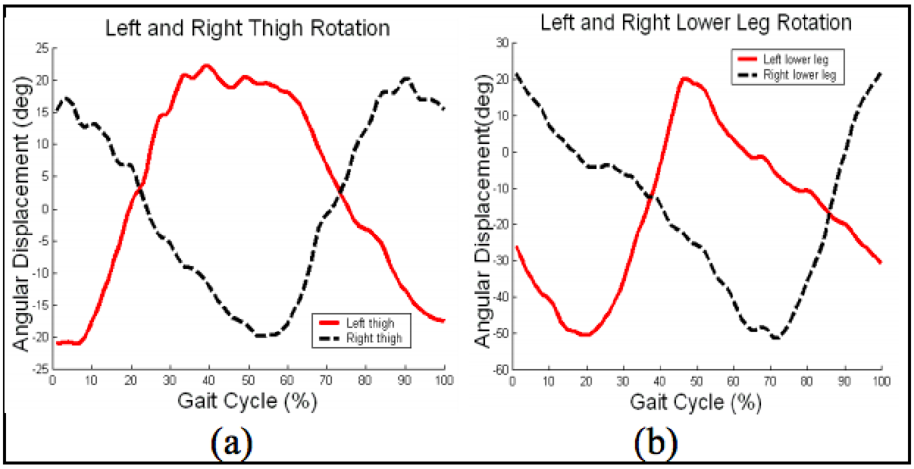

Figure 6: Plots showing a half-period shift in the angular displacement of the left and right thighs (a) and shanks (b) [20]

At Seismic, movement data is obtained from a single trunk-mounted IMU, so it is not possible to take the cross-correlation of the two leg signals. However, the acceleration signal still exhibits the same periodicity as the leg ones. An analogous procedure would then be to take the auto-correlation of the acceleration signal[24].

Variability

Like smoothness in the FRT, variability is a metric that can be applied to many motions. Ideally, movements that are either cyclic (like walking) or repeated (like a golf swing) exhibit little variability. The effects and interpretation of increased variability, though, depend on the movement being examined, the parameters chosen, and the number of repetitions included in an analysis. For instance, increased variability in walking is an indicator of increased fall risk. Increased variability in the kinematics of a basketball free throw, however, may indicate inexperience. In general, when examining variability, it is important to look at the average, the standard deviation, and the time course of the variability.

Variability for individuals is quantified using the average and the standard deviation across repetitions of the same movement. For this analysis, the parameter to be averaged depends on the movement being examined. Commonly chosen parameters include position (or its derivatives) vs time, position vs power, force vs time, and maximum force per repetition. Regardless of the chosen factors, the greater the standard deviation, the greater the variability. Variability can be visualized by overlaying plots of multiple repetitions of the same movement or by plotting the average and the standard deviation, as seen in Figure 7. The amount of variability that can be considered "bad" depends on the movement. For walking, it is possible to use variability of certain gait parameters to distinguish between fallers and non-fallers. For example, Svoboda et al. [21] found that a 0.85 mm (0.033 in) variability of M-L COP

[24] Either the A-P component or the longitudinal component as the M-L component is, in theory for symmetrical walking, equal but opposite for the left vs right legs

displacement during the preswing phase of stance was indicative of increased fall risk.

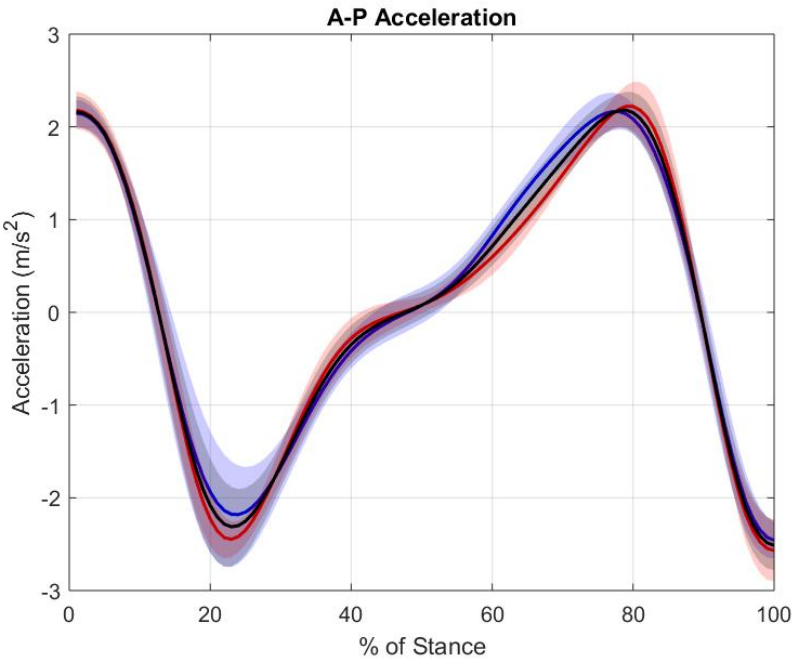

Figure 7: Sample subject data showing the average trunk acceleration in the A-P direction over a single step during the 10 MWT. The shaded areas show standard deviations, while the lines show averages. The black line is the average of every step during the test, the red line is the average of right steps, and the blue is the average of left steps. This person displays highly symmetric walking in the A-P direction and little variability between steps.

It is important also to account for the way variability develops over time (or repetitions). High variability between each repetition of a movement simply indicates poor performance, but high variability after many repetitions is an indicator of fatigue. In this way, endurance may be quantified as the number of repetitions of a movement that can be performed before a threshold value for deviation from the mean is reached. Alternately, it is possible that the early repetitions of a movement vary widely but then stabilize around a certain value after many repetitions. Changing variability is then a sign of motor adaptation and the rate of motor learning can then be quantified as the slope of the measured variable vs repetitions before a stable value is reached. It can be difficult, especially for time-series data, to visualize how movements vary over time or with each repetition in post-processing. For a gross comparison of time-dependent data[25], video footage may be used. It may also be possible to use animated plotting tools to plot and overlay the data from each repetition one-by-one. For analyses looking at a single value per repetition, these may simply be plotted as the value vs repetition number.

[25]For instance, like the acceleration vs percent of stance data seen in Figure 7.

Future Considerations

The tests and metrics discussed above were are all chosen primarily because they can be performed with the equipment currently available on-site at Seismic's headquarters in Menlo Park, CA. Though meeting this requirement means that many of these metrics may also be obtained in situ at Seismic's off-site user testing locations or wherever Seismic products may be used, this also means they may lack the sensitivity and accuracy of the "gold standards" of human movement analysis. Some may also require additional research in an academic setting for further validation against other tried-and-true measures.

With this in mind, future investment in biomechanics equipment such as a motion capture system, force plates, an electromyography system, a treadmill, and O_2 and CO_2 analyzers would greatly expand the analyses that could be performed and the robustness of the measures currently being collected. It would allow for the assessment of movements which may be highly valued by users but not well-documented in academia, such as those performed in an industrial context or during common household activities that do not fall under the rehabilitation community's umbrella of activities of daily living[26]. Furthermore, it would enable Seismic to assess the physiological effects of their products[27].

More immediately, a vast improvement in the robustness of the measures can be achieved by improving the IMU system used. Currently, most of the pilot studies were conducted with a single IMU tethered to a laptop. In the best case scenario, the IMU was attached directly to the skin with additional tape used to secure the USB tether as well. In most situations, the IMU was placed over clothing[28], introducing artifacts from both clothing movement and movement of the cable (particularly in the M-L direction)[29]. The development of a wireless IMU system would greatly reduce these artifacts and may also enable simultaneous data collection from multiple IMUs which would be particularly useful when examining walking.

Finally, more user data should be collected–particularly for the more novel methods discussed here. Doing so provides further internal validation of the methods[30], enables the creation of normative datasets that individual data may be compared to[31], and allows for the creation of more performance ratings beyond the binary "good" or "bad". This is of prime importance for the users with DMD, as many of these methods have never been used within that population and very little data exists for which to judge their movement quality.

[26] This includes only bathing, grooming, toileting, dressing, self-feeding, and transferring and not other common activities like cooking and child-rearing

[27] Such as changes in muscle activation

[28] This was done either because of user sensitivity or because of time constraints.

[29] In addition to the artifacts from the movement of skin and adipose tissue relative to muscles and bones

[30] Which is important in addition to validation via academia because the conditions in which Seismic collects data will never be as controlled or sterile as those found in research

[31] Especially considering research tends to focus more on pathological vs non-pathological movement and does not usually distinguish between shades of "good" movement (outside of sports research).

Conclusions

It is important to note that none of these metrics are meant to be used as a single, definitive measure of movement quality or stability. Movement quality is multi-faceted and complex, with many factors affecting it. The assessments and metrics described were chosen because they address one of six target areas– balance, endurance, trunk stability, hip function, reach quality, or walking quality–, meet the requirements listed above, and can currently be implemented at Seismic. As the capabilities of Seismic's biomechanics team changes, more tools may become available that can provide additional insight into how Seismic technology alters, supports, and improves human performance.

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Appendices

Glossary of Human Movement Terms

- **Acromion process**: the bony process of the scapula. It is a continuation of the scapular spine.
- **Anterior-posterior**: in reference to body planes, movement along or within the sagittal plane, front-to-back.
- **Base of support**: the region defined by the outside perimeter of any body part or assistive devices in contact with the ground or other support structure. In postural studies, this is usually the trapezoidal area defined by the outside perimeter of the feet.
- **Center of gravity**: the point about which the distribution of weight of a body is assumed to be symmetrical. In an average man standing in the anatomical positions, this is located slightly above the navel. For women, it is below the navel.
- **Center of pressure**: the point of application of the ground reaction force.

- **Cervical**: refers to the uppermost section of the spine, consisting of 8 vertebrae.
- **Contralateral**: the opposite side of the body.
- **Duchenne muscular dystrophy**: a genetic disorder, typically affecting boys, characterized by progressive muscle weakness and degeneration due to the absence of the protein dystrophin.
- **Ground reaction force**: the reaction force from the ground, exerted on a body in contact with it.
- **Inertial motion unit**: an electronic device that measures and records an object's orientation through the use of an accelerometer, a gyroscope, and often a magnetometer
- **Limits of stability**: the specific boundaries of space where the body can maintain its position without changing its base of support. In adults, this is 12.5° in the anterior-posterior direction and 16° in the medio-lateral direction when the feet are 4 in apart.
- **Lumbar**: refers to the section of the spine below the thoracic vertebrae, corresponding to the lower back. It consists of 5 vertebrae.
- **Medio-lateral**: in reference to body planes, movement along or within the frontal plane, side-to-side.
- **Minimally clinically important difference**: the smallest difference considered beneficial by a patient.
- **Postural stability**: the ability to maintain the position of the body and the center of gravity within specific limits of stability.
- **Postural sway**: the horizontal movement of the body's center of gravity.
- **Performance of upper limb test**: a test consisting of 22 items used to assess upper limb function in Duchenne muscular dystrophy
- **Proprioception**: one's sense of the position of one's own body parts
- **Sacral**: refers to the lowest portion of the spine, above the coccyx. It consists of 5 vertebrae.
- **Stabilogram**: a plot showing the movement of the body's center of pressure during quiet standing.
- **Thoracic**: refers to the middle section of the spine below the cervical vertebrae in the area of the mid-back. It consists of 12 vertebrae.

Functional Reach Test Protocol

Goal: To quantify static stability using the functional reach test

Equipment:

- 3-axis accelerometer/IMU
- Velcro/tape to attach accelerometer
- Tape to attach yardstick to wall
- Yardstick
- Chair/Yoga ball (optional)

Set-up:

- Yardstick should be taped to the wall at the level of the user's acromion process (boney tip of the shoulder joint above the humeral head)
- Attach IMU to user's back, preferably on the upper thoracic spine (between the shoulder blades)
- Ensure IMU is not located on either scapulae
- If a chair/yoga ball is being used, it should be placed on a hard, level surface near a wall
- User should be able to sit with his/her knees in 90 deg of flexion and feet flat on the ground
- User should not be touching the wall

Recorded Measurements:

- Acceleration data
- Distance 3rd metacarpal head/acromion moves

Basic Test Procedure:

1. Have the user stand next to but not touching the wall.
2. Instruct him/her to lift the arm closest to the wall to 90 deg of shoulder flexion/abduction.
3. Record the position of the 3rd metacarpal head as the starting position. If the user cannot lift their arm to the required 90 deg, note this and record the position of the acromion process as the starting position instead.
4. Have the user reach as far as possible parallel to the wall without taking a step forward.
5. Record the end position of the 3rd metacarpal head/acromion.
6. Repeat the procedure for reaching forward, to the left, and to the right.
7. For each direction, perform 2 trials– a practice trial and an actual trial.

Modified (Sitting) Test Procedure:

1. Have the user sit next to but not touching the wall.
2. Instruct him/her to lift the arm closest to the wall to 90 deg of shoulder flexion/abduction.

3. Record the position of the 3rd metacarpal head as the starting position. If the user cannot lift their arm to the required 90 deg, note this and record the position of the acromion process as the start position instead.
4. Have the user reach as far as possible parallel to the wall without rotating or shifting within the chair or on the yoga ball.
5. Record the end position of the 3rd metacarpal head/acromion.
6. Repeat the procedure for reaching forward (left and right), to the left (side), and to the right (side). For each direction, perform 2 trials– a practice trial and an actual trial.

Sway Test Protocol

Goal: To quantify stability using IMU data collected during quiet standing

Equipment:

- 3-axis accelerometer/IMU
- Velcro/tape to attach accelerometer
- Measuring tape
- Timer
- Balance pad (optional)
- Wheelchair/Yoga ball (optional)

Set-up:

- Test should be conducted on a hard, level surface
- If user is asked to stand on a compliant pad, height of the pad should be recorded
- If user is asked to stand with his/her eyes closed, this should be recorded
- User should not be wearing any shoes; socks are ok
- Attach IMU to user's back
    - Placement is arbitrary, but should be no lower than L4 to ensure trunk movement is captured, not hip/lower extremity
- If user is unable to stand but can be transferred from his/her wheelchair, a yoga ball and the modified procedure may be used
- If user cannot be transferred from his/her wheelchair, the modified procedure may be used with the person sitting away from the back of the chair
- When the modified procedure is used, it may be easier when recording the height of the accelerometer to record first the height of the chair/ball and then add the height of the accelerometer from the seat

Recorded Measurements:

- Acceleration data
- Height of accelerometer from ground
- Height of any pad user is standing on

Basic Test Procedure:

1. Ask user to stand straight with feet shoulder width apart
2. Measure and record height of accelerometer from the ground
3. Allow user to relax for a moment

4. Instruct him/her to stand straight again with feet shoulder width apart and hands relaxed at his/her side. Eyes may either be looking forward or shut
5. Have him/her stand as still as possible for 30-60 s
6. Have user relax

Modified (Sitting) Test Procedure:

1. Ask the user to sit up straight
2. Measure and record height of accelerometer from the ground
3. Allow user to relax for a moment
4. Instruct him/her to sit up straight again with hands relaxed at his/her side. Eyes may either be looking forward or shut
5. Have him/her sit as still as possible for 30-60 s
6. Have user relax

10 Meter Walk Test Protocol

Goal: To quantify functional ability and stability and movement quality during walking

Equipment:

- Timer
- Velcro/tape to attach IMU
- Tape to mark path
- IMU

Set-up:

- Mark out a straight 10 m path over an even surface
- Attach IMU to user's back, preferably on the upper thoracic spine (between the shoulder blades)
- Ensure IMU is not located on either scapulae

Recorded Measurements:

- Acceleration data
- Time to walk 10 m
- Preferred vs maximal speed

Basic Test Procedure:

1. Have the user stand at the start of the path.
2. Instruct him/her to walk normally down the path at either his/her preferred pace or as fast as safely possible. The user should continue walking for a few steps after reaching the end of the path. Record the pace used.
3. Have the user walk down the path while someone times them.
4. Record the starting leg (first leg to strike within the path).
5. Stop the timer when he/she crosses the 10 m mark.
6. Record the total time.

Using a Trunk-Mounted Accelerometer to Identify Gait Events

A novel method of identifying gait events was developed to provide further insight into users' gait quality. This method utilizes a single, trunk-mounted IMU, circumventing the need for a motion capture system or force plate. It was developed based on work done by Meuleman et. al [24] which utilized a Vicon optical motion capture system. Preliminary validation was done by comparing the number of steps identified via this method to the number of steps visually identified from videos of subjects performing the 10 Meter Walk Test and to the number of steps identified during a long walk. Well-documented characteristics of gait such as the relative duration of stance and swing and the displacement of the COM provided further reality checks.

The method appears to have some efficacy, at least for non-pathological gait, in determining step count and estimating the timing of gait events. Additional validation is still needed to quantify the temporal error associated with this method.

Procedural Notes:

- Events are identified using the A-P component of the trunk acceleration vector, corresponding to the forward direction of walking (Figure 8).
- Local minima correspond to toe-offs[32].
- Local maxima correspond to footstrikes[33].
- The left or right leg can be identified either from video, if recorded, or from tester notes on which foot struck first.
- Event timestamps from the A-P component can then be used to identify gait events in the M-L and vertical directions.

[32] This is an estimation as toe-off typically occurs a little after the A-P acceleration of the COM reaches a local minimum.

[33] This is also an estimation. The timing of the footstrike in relation to a local maximum may vary between heel-strikers, midfoot-strikers, and toe-strikers.

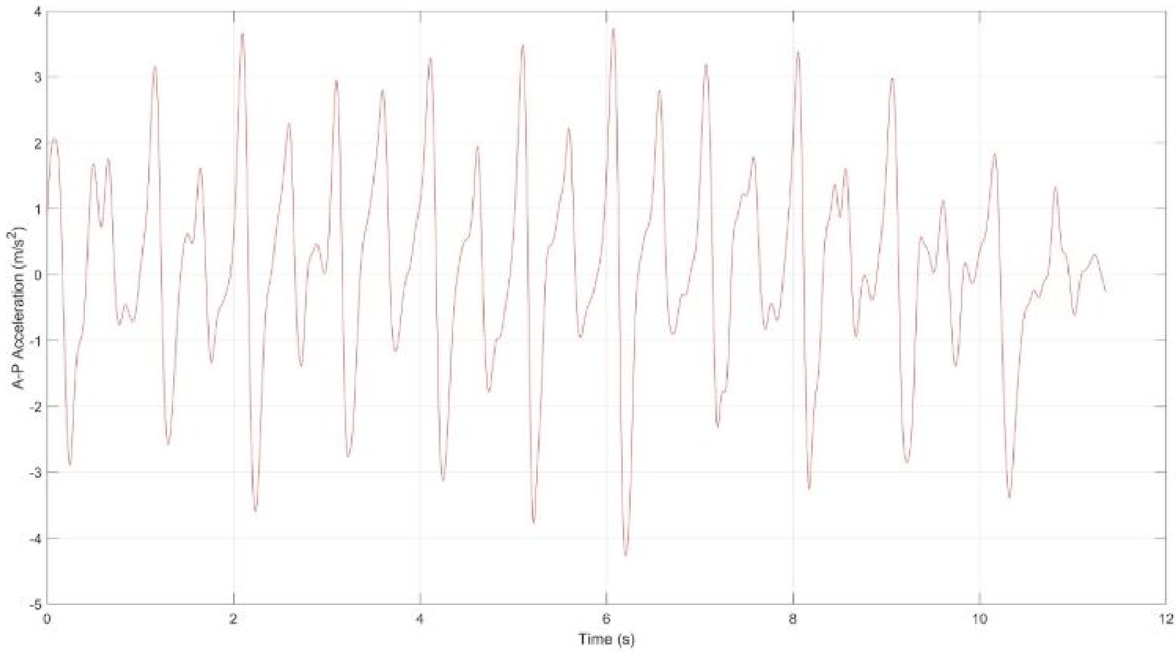

Figure 8: Sample user data from a boy with DMD showing the trunk acceleration in the A-P direction during the 10 MWT. Note, the values shown here have not been degravitized but should be for symmetry and variability analyses. The data also displays an asymmetry between alternating legs, which is in agreement with previous observations from a physical therapist.

Reality Checks:

- The number of identified steps matches the number seen in any corresponding video.
- Any missteps or gait disturbances appear within the data as variability in the affected steps.
- The duration of swing phase is approximately 40% of the gait cycle [22].
- The duration of stance phase is approximately 60% of the gait cycle [22].
- Each period of double support[34] is approximately 10% of the gait cycle [22].
- The duration of single support is approximately 40% of the gait cycle [22].
- The duration of stance decreases with increasing velocity.
- The vertical trunk acceleration rises and then falls during swing phase as the body COM is propelled upward after toe-off and then falls back down during terminal swing.
- Noted unilateral conditions such as uneven leg strength or injuries or surgeries affecting one leg, may be evidenced by temporal asymmetries or differences in the average magnitude of acceleration between each leg.

[34]There are two during stance phase, initial double support and terminal double support.

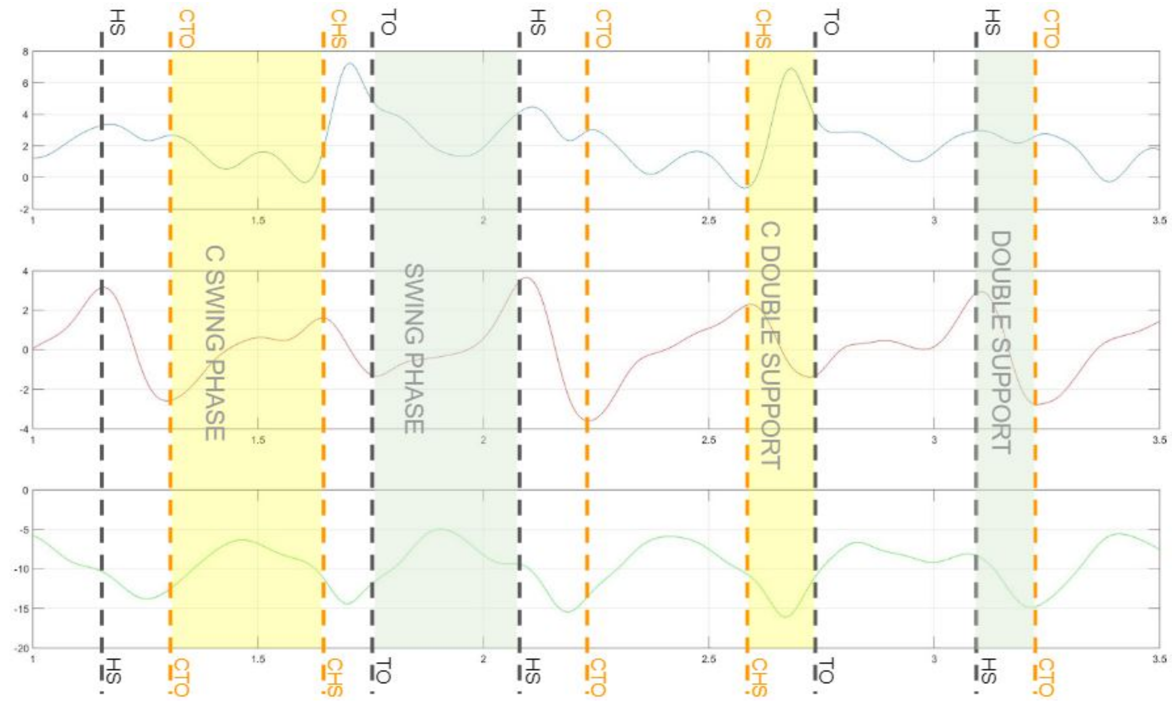

Figure 9: The same data from Figure 8 (middle plot) zoomed in to show 5 steps. The top plot shows the M-L component of the acceleration for the corresponding time period, while the bottom shows the vertical component. Heel strike (HS), toe-off (TO), contralateral heel strike (CHS), and contralateral toe-off (CTO) are marked with the dashed lines. The shaded areas show swing phase and double support. Though unlabeled, stance phase can be identified as the area between one HS and the subsequent TO of the same foot. It encompasses two periods of double support and the swing phase of the contralateral foot. The plots show the expected durations for each phase of the gait cycle and the characteristic rise and fall of the trunk vertical acceleration during swing.

Who We Are

Seismic

Seismic is a start-up specializing in the nascent domain of powered clothing. Their products are a fusion of apparel and robotics that aim to discreetly provide users with extra strength and support through key movements. Though similar in purpose to traditional exoskeletons, Seismic's powered garments lack rigid links and do not provide assistive torques at joints of interest, instead providing assistive forces across joints, similar to the natural function of muscle-tendon units. Their current product line includes a core wellness suit targeted at active, older adults and a suite of rehabilitative garments for boys with Duchenne muscular dystrophy which is still in development.

Originally, founded as SuperFlex, they are a spin-off of the Stanford Research Institute and DARPA's Warrior Web Project. They are located at the heart of Silicon Valley in sunny Menlo Park, California in the United States.

The Intern

Officially, my title at Seismic was Research Engineer Intern and I worked within the biomechanics team. My primary task, described above, was to find ways to quantify the effects and efficacy of Seismic's technology by quantifying movement quality. I was given a single IMU, tethered to a laptop via a USB cable and a lot of autonomy to accomplish this. When needed, I sought guidance from my supervisors (themselves mechanical engineers), the software team, and our resident physical therapist, but I primarily worked alone unless I was experimenting on my coworkers.

Though it was too late in the development of the core wellness suit to make changes based on my findings and too early in the development of the DMD products to make definitive conclusions (because only a small number of boys had been brought in for user testing at the time of my departure), the graphs of my results were still used to further illustrate the capabilities of Seismic's products to current and future investors and the methods I proposed are being used in current user testing scenarios.

I was also given smaller, related tasks such as evaluating new IMU systems, researching and selecting equipment for a new biomechanics lab, and reporting data anomalies to either the software team or the user testing team (at the time, I was the only person looking at the user data we were collecting) so they could be fixed. Additionally, in the course of completing my primary assignment, I completed "debugging" tasks like quantifying the thermal drift in the suit-mounted IMUs and examining the effect of motor vibration on the gyroscope reading.

Unofficially, with my experience looking at biomechanics data and coding skills, I was the resident data scientist. In this capacity, I did work programming motor trajectories, determining motor timing based on user testing data, designing new motor requirements based on the same data, and comparing movements performed within the suit to data from a collaborator at a biomechanics lab.

Additionally, given my experience with data collection and human subject testing, I functioned as an extra member of the user testing team when off-site user tests began and as a member of the development team for the Duchenne products when the boys with DMD would come in to test the prototypes.

Overall, my role within the company was a flexible one. I was hired to exam human movement and shed light on how their products actually affect users, but given my skillset and Seismic's small size was often asked to fill in where there were holes in capabilities.