Textile band electrodes as an alternative to spot Ag/AgCl electrodes for calf bioimpedance measurements

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Abstract

Objective: To evaluate the performance of five different types of textiles as band electrodes for calf bioimpedance measurements in comparison with conventional spot Ag/AgCl electrodes. Approach: Calf bioimpedance measurements were performed in 10 healthy volunteers with five different textile materials cut into bands and Ag/AgCl spot electrodes as a baseline. Collected bioimpedance data were analyzed in terms of precision, fit error and presence of measurement artifacts. Each textile material was also evaluated for participant comfort. Main Results: Bioimpedance values for spot electrodes were higher at low frequencies as compared with band electrodes but not at high frequencies. This suggests that spot electrodes have frequency dependent current distributions that adversely impact their use for volume measurements and band electrodes are preferable. The SMP130T-B fabric had the highest precision and the lowest best fit error to the Cole model of the tested textile materials. However, it was the least comfortable textile and most expensive. The Stretch material performed slightly worse than the SMP130T-B fabric, but was half the cost and the most comfortable. Significance: These results suggest that there are suitable textile materials for use as dry, band electrodes for calf bioimpedance measurements and that these band electrodes enable greater current uniformity. These textiles could be integrated into a compression sock for remote monitoring of diseases such as Congestive Heart Failure.

1. Introduction

Bioimpedance is a measurement of the electrical properties of tissue typically obtained by driving a small current through the body and sensing the resulting voltage [1, 2]. The technique can be used for a number of different applications, including body composition [3], hemodynamic monitoring [4], and assessment of volume status [5, 6]. Most bioimpedance measurements are 'tetrapolar' and involve the placement of two pairs of electrodes; outer electrodes that drive current and inner electrodes that sense the resulting voltage. A tetrapolar configuration minimizes the impact of electrode polarization at frequencies above about 1kHz assuming equal electrode area and material [7]. These electrodes are most frequently spot electrodes such as those used for ECG measurements.

Bioimpedance measurements for full body composition involve placement of electrodes on the wrist and ankle and measure the properties of the arm, trunk and leg in series. In recent years there has also been an interest in 'segmental' bioimpedance that involves placing electrodes closer together to measure just one smaller area of the body. These measurements can be used, for example, for monitoring knee or ankle joint health [8, 9], for monitoring fluid status during hemodialysis using calf bioimpedance measurements [10–13] or in Congestive Heart Failure patients [14, 15].

When a wrist to ankle bioimpedance measurement is performed, the long conductor length of the measurement ensures that the current is distributed evenly through the measured segments for the majority of the current path, other than small areas of constriction close to the electrodes [16]. However, segmental measurements involve shorter inter-electrode spacing and
the resulting current does not distribute evenly throughout the tissue [17]. This uneven current distribution results in the majority of current flowing through the tissue close to the electrode sites, which prevents accurate volume estimation. Simulations from [17] suggest that band electrodes that wrap around the calf could improve current distribution as current is injected more uniformly around the circumference of the calf.

In addition to issues with current distribution, Ag/AgCl spot electrodes also require sticky adhesives and dry out over time. In order to implement a band electrode and also allow for long term use, appropriate dry electrode materials must be considered. There have been a number of studies evaluating conductive textiles, polymers and metals for bioimpedance measurements [18–22]. Textiles have the advantage of being readily integrated into bands and garments. However, two major challenges in using textiles for bioimpedance measurements are high skin-electrode interface impedance [18] and the need to secure the textiles in contact with the skin to prevent motion artifacts [23]. Fortunately, using band electrodes may address both of these issues as the higher contact area should reduce skin-electrode contact interface impedance and applying pressure, for example using a compression garment, could help mitigate motion artifacts.

The purpose of this study was to evaluate several commercially available textile-based dry band electrodes in healthy young adults in comparison with conventional Ag/AgCl spot electrodes to (1) select a material to integrate into a compression sock for remote calf bioimpedance monitoring and (2) compare measurement results from spot and band electrodes to determine which is more suitable for calf volume estimation. Calf bioimpedance measurements were performed in healthy volunteers and different metrics of performance such as measurement-to-measurement variability, measurement artifacts, and participant comfort were measured. A successful textile material should have comparable measurement results with improved comfort compared with Ag/AgCl electrodes.

2. Methods

2.1. Informed consent

All participants gave informed consent to participate in the study and all experimental procedures were approved by the Swarthmore College Institutional Review Board (Protocol # 17-18-044). Inclusion criteria for the study were: over the age of 18 and not currently enrolled as a student of the research supervisor. Exclusion criteria for the study were: implanted medical devices, skin sensitivity to medical adhesives and electrodes, pregnancy, amputations, metal in legs, and ulcers or other skin conditions at the potential electrode sites.

2.2. Textile materials

Five different textile materials were selected for evaluation based on materials that show promise from previous research involving bioimpedance measurements with textiles (see e.g. [18, 24]). In particular, textiles that were nylon and silver based have performed well in previous studies. Additionally, woven fabrics have had lower contact impedance than knitted fabrics. The characteristics and descriptions of the textile materials are presented in tables 1 and 2. All fabrics were purchased from Marktek Inc. (Chesterfield, MO, USA), except for the Stretch material, which as purchased from Less EMF Inc. (Latham, NY, USA).

Table 1. Properties of the different textile materials used as electrodes in the present study.

| #  | Name         | Surface Resistivity Ω/sq | Thickness (mm) | Basis Weight (g/m²) | Stretch          |
|----|--------------|--------------------------|----------------|---------------------|-----------------|
| 01 | SMP130-B     | < 100                    | 0.45 +/- 10%   | 142 +/- 10%        | up to 95%       |
| 02 | Stretch      | < 1                      | 0.4            | 146                 | 100% len., 65% wid. |
| 03 | SMP130T-B    | < 3                      | 0.45 +/- 10%   | 157 +/- 10%        | up to 95%       |
| 04 | SBRM48       | < 0.3                    | 0.09 +/- 12%   | 43 +/- 10%         | N/A             |
| 05 | SBAL317      | < 1                      | 0.26 +/- 10%   | 62 +/- 10%         | N/A             |

Table 2. Manufacturer descriptions of textile materials used as electrodes in the present study.

| #  | Name        | Description                                                                 |
|----|-------------|-----------------------------------------------------------------------------|
| 01 | SMP130-B    | Silverized Elastic Knit, nylon/spandex knit fabric coated with 99% pure silver and then thin polyurethane protective layer, dark grey colored |
| 02 | Stretch     | Silver coated on fabric made up of 76% nylon and 24% elastic material, dark grey colored and stretchy |
| 03 | SMP130T-B   | Silverized Elastic Knit, highly conductive, nylon/spandex, elastic, knit fabric coated with 99% pure silver plus a thin, protective polyurethane, dark grey colored |
| 04 | SBRM48      | Nylon ripstop fabric coated with 99.9% silver |
| 05 | SBAL317     | Nylon tricot knit fabric coated with 99.9% silver, designed with high ionic Ag release, especially for antimicrobial and wound care products in mind |
2.3. Study design

After consenting to participate in the study, participants were instrumented first with Ag/AgCl (3M-2560) electrodes as a reference, followed by the five textile materials in the same order each time. Each of the five textile materials were cut into 4 cm wide strips with Velcro on either end to wrap snugly around the participant’s calf (see figure 1). Bands of lengths ranging from 35 cm to 45 cm were made to ensure a good fit for each participant as needed and the Velcro on each end of the strip allowed for fine adjustment of the band fit. Medical tape was then wrapped around the textile bands to secure the textiles in place and mimic the pressure from the eventual compression sock implementation. Conductive thread was used to connect the textile electrodes to the measurement system (ImpediMed SFB7, ImpediMed Inc., Carlsbad, CA). All electrodes were placed longitudinally along the calf (in the case of the Ag/AgCl electrodes, they were placed on the lateral side of the calf [10, 12]). The inner, voltage electrodes were spaced 10 cm apart and spaced around the mid-point between the middle of the patella (knee) and lateral malleolus (ankle). The outer, current electrodes were spaced 5 cm on either side of the voltage electrodes. Calf circumference measurements were performed at the point of the largest calf circumference. Calf hair was qualitatively evaluated by the researchers on a scale of 1 (no hair) to 5 (very hairy).

Calf bioimpedance measurements were performed in a sweep of 256 logarithmically spaced frequencies from 3 kHz to 1 MHz up to six times each using the commercial bioimpedance device. Measurements were performed with the participant in a sitting position with their upper and lower legs at right angles and feet flat on the floor. Participants were able to freely stretch their limbs between measurements. The entire study duration was about 1 h.

2.4. Bioimpedance spectroscopy measurements

Bioimpedance measurements performed across frequencies are known as bioimpedance spectroscopy (BIS) measurements. These measurements can be analyzed at individual frequencies or fit to a model. The most frequently used model for BIS measurements is the Cole model [25]. The Cole model is a four-parameter model where the measured bioimpedance $Z$ can be represented by

$$Z = R_\infty + \frac{R_0 - R_\infty}{1 + (j\omega\tau)^a}$$  \hspace{1cm} (1)

where $R_0$ and $R_\infty$ are the resistance at frequencies $\ll 1/\tau$ and $\gg 1/\tau$, respectively. $\tau$ is the characteristic
Table 3. Functional Requirements and their associated parameters for evaluating the performance of each textile material as a dry electrode.

| Requirement                  | Parameter                                      |
|------------------------------|------------------------------------------------|
| Usable Data                  | 99%+ measurement runs conform to Cole model    |
| High precision within consecutive bioimpedance sweeps | 1% run to run                                 |
| Low Cole Model Fit Error     | mean 1% across frequencies                     |
| Participant Comfort          | More comfortable than Ag/AgCl                  |

The magnitude of each bioimpedance measurement from 4.7 kHz to 200 kHz was fit to the Cole Model using MATLAB’s lscurvefit algorithm. This frequency range was selected to minimize the influence of low and high frequency artifacts (see section 4.5). The initial conditions for each fit were iterated 50 times using MATLAB’s MultiStart algorithm to avoid local minima. After the fit was performed, the Cole parameters $R_0$ and $R_{\infty}$ were analyzed to evaluate textile performance.

2.5. Evaluating textile performance

There are several requirements for a successful textile material. The material should have high precision run to run, the material should obtain measurements with low fit error to the Cole model (i.e. there should be minimal measurement artifacts and measurement noise), and it should be comfortable for participants to wear (see table 3).

The precision $p$ is defined as

$$p(S, t) = 100 \times \frac{\text{std}(R(S, t))}{\text{mean}(R(S, t))}$$

where $R$ is the set of measurements performed for a given participant $S$ and textile $t$ (in this case either the Cole parameter $R_0$ or $R_{\infty}$). The precision $p$ is therefore the percentage variation of the different measurement runs, allowing comparisons across participants and textiles.

The Cole model fit error was calculated as the percent difference between the fitted data and the raw data, the result of which was averaged across all frequencies and participants to return a single value per textile. The precision and fit error were only calculated in cases where the data were able to be fitted to the Cole model (e.g. no parameter hit an upper or lower bound). Any data that hit a parameter bound was considered ‘unusable’ and excluded from further analysis; any data that had deviations to the Cole model (e.g. at low or high frequency) were considered artifacts.

Comfort of each electrode was evaluated using a survey that the participant completed throughout testing using a 1–5 scale where 1 is least comfortable and 5 is most comfortable. These data were subtracted by the Ag/AgCl score for comparison between participants.

2.6. Evaluating textile properties

To better understand the bioimpedance measurement results in healthy participants, measurements were also performed using the same five textiles and Ag/AgCl electrodes on an Agar-Agar phantom. Measurements were performed using a phantom, rather than human skin, because skin impedance can change by up to 20% over the course of 2 h [18]. The phantom was shaped in a block similar to that used by Beckmann et al [18]. Using a gold electrode on the bottom of the phantom with a small patch of textile on the top, the impedance between the gold electrode and the textile patch was measured using an Agilent E4980A Precision LCR meter from 20 Hz to 1 MHz. The phantom was allowed to rest until the surface reached room temperature, and measurements were performed in quick succession to minimize the impact of any changes to the electrical properties of the phantom.

3. Results

3.1. Patient demographics

Demographic information for participants in the study ($N = 10$; five female) is presented in table 4. Participants were 22.4 ± 2.9 years old, weighed 69.5 ± 18.3 kg, were 173.6 ± 12.3 cm tall, and had calf circumferences of 38.9 ± 6.7 cm. Eight out of ten patients had no or very little calf hair (scores of 1 or 2); one participant had moderate calf hair (score of 3) and one participant had a large amount of calf hair (score of 5).

3.2. Cole parameters

Mean values for the Cole parameters $R_0$ and $R_{\infty}$ are presented in tables 5 and 6, respectively. In some cases (represented by the ‘–’ in the cell), data were excluded from subsequent analysis because the result of the fitting algorithm pinned said parameter to an upper
Table 4. Participant Demographics (N = 10; five female). Calf circumference was measured at the widest part of the calf.
Calf hair was scored on a scale from 1 (no hair) to 5 (very hairy).

| Participant | Age (yrs) | Gender | Weight (kg) | Height (cm) | Calf Circ. (cm) | Calf Hair |
|-------------|-----------|--------|-------------|-------------|----------------|-----------|
| S000        | 22        | M      | 97.5        | 185         | 43             | 1         |
| S001        | 24        | M      | 56.7        | 170         | 37             | 2         |
| S002        | 22        | F      | 56.7        | 167         | 35             | 1         |
| S003        | 22        | M      | 7.4         | 77          | 33             | 2         |
| S004        | 21        | F      | 54.4        | 168         | 34             | 1         |
| S005        | 22        | M      | 78.9        | 183         | 47             | 3         |
| S006        | 20        | M      | 104.3       | 198         | 53             | 5         |
| S007        | 21        | F      | 62.6        | 168         | 38             | 2         |
| S008        | 20        | F      | 54.5        | 165         | 36             | 1         |
| S009        | 30        | F      | 59          | 155         | 33             | 2         |
| Mean ± SD   | 22.4 ± 2.9| 69.5 ± 18.3| 173.6 ± 12.3| 38.9 ± 6.7 | 2 ± 1.2       |

Table 5. Mean R∞ values for each participant and textile. Data was missing from S003 t02 due to a file transfer error. The dashes (-) indicate that the parameter was set either to the lower or upper bound (i.e. there was no good fit for the data) and these data are therefore excluded from analysis.

| Participant | Ag/AgCl | t01 | t02 | t03 | t04 | t05 |
|-------------|---------|-----|-----|-----|-----|-----|
| S000        | 40.17   | 27.10 | 29.10 | 32.78 | —   | 29.50 |
| S001        | 37.05   | 19.87 | 16.38 | 17.96 | —   | 18.99 |
| S002        | 51.94   | 32.54 | 36.94 | 29.32 | 33.69 | 36.51 |
| S003        | 46.83   | 33.20 | N/A  | 28.64 | —   | 32.37 |
| S004        | 46.03   | 31.44 | 28.98 | 30.59 | —   | 27.37 |
| S005        | 38.78   | 16.12 | 24.23 | 21.66 | 24.43 | 19.99 |
| S006        | 29.76   | 20.53 | 19.01 | 20.64 | —   | 20.62 |
| S007        | 34.50   | 20.39 | 22.37 | 20.86 | —   | —     |
| S008        | 45.48   | 27.63 | 26.04 | 28.33 | —   | —     |
| S009        | 45.83   | 28.04 | 25.41 | 28.11 | —   | 32.28 |
| Mean ± SD   | 41.64 ± 6.74 | 25.71 ± 6.02 | 25.38 ± 6.05 | 25.89 ± 5.09 | 29.06 ± 6.55 | 27.20 ± 6.62 |

Table 6. Mean R∞ values for each participant and textile. Data was missing from S003 t02 due to a file transfer error. The dashes (-) indicate that the parameter was set either to the lower or upper bound (i.e. there was no good fit for the data) and these data are therefore excluded from analysis.

| Participant | Ag/AgCl | t01 | t02 | t03 | t04 | t05 |
|-------------|---------|-----|-----|-----|-----|-----|
| S000        | 15.05   | 16.76 | 16.51 | 17.32 | —   | 17.23 |
| S001        | 13.98   | 12.79 | 10.17 | 10.33 | 17.48 | 10.51 |
| S002        | 25.47   | 21.63 | 24.65 | 19.75 | 21.91 | 23.83 |
| S003        | 17.82   | 19.39 | N/A  | 17.93 | —   | 20.16 |
| S004        | 18.09   | 15.83 | 17.41 | 18.85 | 61.86 | 18.46 |
| S005        | 12.01   | —     | 11.92 | 12.21 | 14.80 | 11.93 |
| S006        | 12.49   | 12.89 | 10.89 | 12.94 | 90.88 | 13.31 |
| S007        | 16.15   | 12.08 | 13.43 | 12.36 | —   | —     |
| S008        | 16.92   | 15.08 | 15.03 | 15.29 | —   | —     |
| S009        | 20.98   | 17.25 | 15.25 | 16.11 | —   | 19.84 |
| Mean ± SD   | 16.90 ± 4.06 | 15.97 ± 3.39 | 15.03 ± 4.38 | 15.31 ± 3.20 | 41.39 ± 33.64 | 16.91 ± 4.6 |

lower bound, indicating that no suitable fit could be found. This occurred most frequently in textile t04, where data from only 2/10 participants was valid for R0, and data from only 5/10 participants were valid for R∞. This also occurred twice for t05 for both R0 and R∞.

Ag/AgCl had statistically significantly higher R0 values than each of the textiles t01, t02, and t03 (paired t-test, all p-values < 0.001). There were no statistically significant different differences for R0 between textiles t01, t02, and t03 (paired t-test, all p-values > 0.05).

For R∞, Ag/AgCl was only statistically significantly higher than t02 (paired t-test, p < 0.05); there were no statistically significant differences between t01/t02/ t03 (paired t-tests, all p-values > 0.05). Figures presenting the Cole parameters R0 and R∞ for Ag/AgCl and textiles t01, t02 and t03 are presented in figures 2 and 3. For R0, Ag/AgCl electrodes consistently had the highest R0 values. The lowest values for R0 varied across participants (t01: 3, t02: 5, t03: 1, 1 excluded due to missing t02 data). For R∞, Ag/AgCl less consistently had the highest R∞ values (5/10
Figure 2. Measured $R_0$ values for Ag/AgCl and textiles t01, t02, and t03 across participants. Each bar represents the mean ± the standard deviation of the six measurements.

Figure 3. Measured $R_\infty$ values for Ag/AgCl and textiles t01, t02, and t03 across participants. Each bar represents the mean ± the standard deviation of the six measurements.

Table 7. Mean Percent Precision for $R_0$ in percent (100 * std($R_0$) / mean($R_0$)) for each participant and textile. The dashes indicate that the data were not suitably fit to the Cole model (e.g. parameter set to the upper or lower bound).

| Participant | Ag/AgCl | t01 | t02 | t03 | t04 | t05 |
|-------------|---------|-----|-----|-----|-----|-----|
| S000        | 0.15    | 0.50| 1.68| 0.64| —   | 0.14|
| S001        | 0.34    | 0.22| 0.25| 0.28| —   | 0.05|
| S002        | 0.06    | 0.10| 0.20| 0.36| 0.72| 0.33|
| S003        | 0.11    | 0.24| —   | 0.96| —   | 1.00|
| S004        | 0.04    | 0.23| 0.19| 0.21| —   | 0.14|
| S005        | 0.73    | 3.51| 0.14| 0.20| 1.18| 0.47|
| S006        | 0.17    | 0.82| 0.29| 0.49| —   | 0.24|
| S007        | 0.09    | 0.10| 0.16| 0.17| —   | —   |
| S008        | 0.11    | 0.18| 0.14| 0.08| —   | —   |
| S009        | 0.06    | 0.31| 0.12| 0.14| —   | 0.09|
| **Mean ± SD** | **0.19 ± 0.21** | **0.62 ± 1.04** | **0.35 ± 0.50** | **0.35 ± 0.27** | **0.95 ± 0.33** | **0.31 ± 0.31** |
participants). The lowest values were also split among participants (t01: 3, t02: 4, t03: 1, t02/t03 tied: 1, 1 excluded due to missing t02 data). When a textile was the lowest in $R_{\infty}$, it was also the lowest for $R_0$ 8/10 times.

3.3. Measurement precision

Data for the measurement precision for each participant and textile is presented in tables 7 and 8 for $R_0$ and $R_{\infty}$, respectively. For $R_0$, precision was less than 1% for all participants for Ag/AgCl and textile t03 and for 9/10 participants for t01 and t02. Textiles t04 and t05 did not score as well: for t04, 8/10 participants did not have usable $R_0$ data, although the data that was usable had precision of 0.72% and 1.18%. Textile t05 was slightly better; there was only unusable data from 2/10 participants, and of the remaining participants, 7/8 had precision data lower than 1%; the remaining participant had a precision of 1.00%.

Precision data for $R_{\infty}$ was similar to the $R_0$ data. Precision was less than 1% for all participants for Ag/AgCl and textiles t02 and t03. One participant for textile t01 did not have valid Cole model data, but all remaining data were usable and had precision < 1%. Data for textiles t04 and t05 had a larger number of participants with usable data compared with $R_0$, but only 2/10 participants had precision < 1% for t04. There were 8/10 participants who had precision < 1% for textile t05.

3.4. Cole model fit error

Normalized mean fit error across frequency and participant was lowest for the Ag/AgCl electrodes and t03 electrodes, although the t03 electrodes had a slightly higher standard deviation in fit error (0.0025 ± 2.4 $e^{-18}\%$ and 0.0025 ± 3.48$e^{-18}\%$, respectively). Textiles t01 and t02 also had low fit errors (0.0075 ± 6.2$e^{-18}\%$ and 0.01 ± 3.24$e^{-17}\%$, respectively), although these fit errors were 3–4 times larger than those of the Ag/AgCl and t03 electrodes. Fit errors for t04 and t05 for participants that had data that had an acceptable fit to the Cole model (i.e. a parameter did not hit the upper/lower bound) were lower than fit errors from t01 and t02, but higher than t03 (0.0037 ± 9.1$e^{-18}\%$ for t04 and 0.004 ± 3.1$e^{-18}\%$ for t05). This included participants S002 and S005 for t04, and all participants except S007 and S008 for t05.

3.5. Participant comfort

Textiles t01, t02, t04, and t05 were statistically significantly rated more comfortable than Ag/AgCl electrodes with comfort scores subtracted by the Ag/AgCl score of 1.5 ± 1.3 (t01), 1.7 ± 1.7 (t02), 1.1 ± 1.1 (t04) and 1.7 ± 1.2 (t05). Textile t03 was not rated more comfortable than Ag/AgCl, with a normalized score of 0.7 ± 2.0.

3.6. Electrode impedance

Testing of the electrode impedance using an Agar-Agar phantom are presented in figure 4. Textile t02 had the lowest contact impedance, with textiles t01 and t03 close behind. Textile t04 had the second highest contact impedance, and textile t05 had the highest contact impedance.

4. Discussion

4.1. Summary of textile material performance

Although textile t03 performed best in terms of measurement quality (usable data, precision, fit error), participants considered it less comfortable than other textiles such as t01 or t02. It was also the most expensive material tested. Textile t02 has the highest scores despite slightly worse performance in terms of data quality, but was less expensive than t01.

4.2. Comparing spot and band electrodes

At low frequencies, Ag/AgCl (i.e. ‘spot’ electrodes) consistently had higher bioimpedance values than measurements using textiles. Previous research has shown that differences in skin-electrode contact area,
such as between ‘spot’ and ‘band’ style electrodes, can impact measured bioimpedance \[16, 17, 20\]. In particular, an increase in skin-electrode contact area (as with the band electrodes in this study) results in decreased measured bioimpedance because current flows more uniformly through tissue, rather than being concentrated near the current electrode sites (as with the spot electrodes in this study).

However, at high frequencies bioimpedance values for Ag/AgCl electrodes were not consistently higher than measurements using textiles. This may be due to the properties of muscle tissue. Muscle tissue is anisotropic due to the long, narrow shape of muscle cells, the myofibrils inside each cell, and the connective tissue in the muscle \[7, 26\]; the measured impedance is lower when current flows parallel to the muscle cells as opposed to perpendicularly. Gómez Sánchez et al showed that muscle anisotropy decreases by 6 times between 1kHz and 1 MHz \[7\]. This would then explain why \( R_\infty \) values for the textile band and Ag/AgCl spot electrodes were more similar than the \( R_0 \) values. We have previously used simulation data to show that measurements at the side or back of the calf tend to result in current distribution localized near the electrodes \[17\]. If one repeats these simulations with the anisotropy of muscle tissue removed, the results for a side electrode placement and a band electrode placement look much more similar, supporting the idea that for spot electrodes current is more uniformly distributed at high frequencies than at low frequencies (see figure 5).

Differences in the spot electrode current distribution at low and high frequencies has negative consequences for any measurement that involves the relationship between calf bioimpedance measurements at both low and high frequencies, including calf volume estimation. Because current at low frequencies flows through a relatively small part of the leg, there will be an underestimation of calf extra-cellular water when using spot electrodes as compared with band electrodes. At high frequencies, current will flow through the calf similarly for both the spot and band electrodes, so these values should be more reflective of the actual calf volume. This would then also lead to an underestimation of the calf extra-cellular water to calf total water ratio (cECW/cTW) or an overestimation of the \( R_0/R_\infty \) impedance ratio for the Ag/AgCl spot electrodes.

### 4.3. Calf bioimpedance correspondence between textile electrode materials

Although there were no statistically significant differences between \( R_0 \) and \( R_\infty \) for textiles t01, t02, and t03 in this cohort, there were differences within participants that sometimes exceeded 5 Ω (see figures 2 and 3). These differences observed are on the same order as those differences observed by other researchers \[19, 22, 27\]. Because the electrode contact surface area for each textile material should be very similar, we hypothesize these differences are due to differences in electrode placement. Movement of the current or voltage electrodes up or down the calf can impact the current distribution through the calf and the measurement volume, respectively. These could alter the measured calf bioimpedance on the order of magnitude of the changes observed here. For example, the measured resistance is proportional to the voltage inter-electrode distance, so changes of 1–2 cm could result in differences of 10%–20%. Although the inter-electrode spacing was marked on the calf during measurements to ensure repeatability textile-to-textile, electrodes may have slid on the calf, causing these discrepancies. Securing the electrodes at the appropriate spacing in the eventual wearable implementation should allow for more consistent measurements. Such an implementation must also ensure robustness.
to translation of the electrodes along the calf, as the location of the electrodes along the calf could shift within and between wears.

4.4. Impact of textile properties on measurements
Textiles t02 and t03 were the best performing textiles with textiles t01 and t05 close behind. Textile t04 performed particularly poorly. Textile t04 was the only woven textile (all others were knit, and t05 was tricot knit). It was also the lightest textile chosen. In their study of the characterization of various textile materials, Beckmann et al. found that the woven textile they tested had the lowest contact impedance, which might suggest that a woven textile would be preferred over a knit material [18]. However, there could be differences between the woven textile in that study and in the present study.

The measured contact impedance seems to relate to the actual measured performance for tetrapolar measurements, with textiles t01 and t03 having lower contact impedance than textiles t04 and t05. Textiles t01 and t03 are similar except their surface resistivity, which does not appear to affect the tetrapolar or contact impedance measurements significantly. The tricot knit of textile t05 is more porous than the knit of textiles t01, t02, t03, which could explain the higher measured contact impedance of textile t05 as there are fewer filaments contacting the skin per unit area. This would be consistent with the results observed in [18].

In the instances where the t04 data were unusable, the measured impedance was very high, consistent with a measurement without a good connection. It is possible that there was not good contact between textile t04 and the skin in our measurements, or between the textile and the conductive thread, possibly due to the light weight of the textile.

4.5. Artifacts in calf bioimpedance measurements
Artifacts in the calf bioimpedance measurements were observed in a subset of participants and textiles. These artifacts occurred both at low frequencies and at high frequencies. The low frequency artifacts were primarily in the phase, where there is positive phase at frequencies over 10 kHz (see figure 6). The magnitude also curves upward at low frequencies up to about 6 kHz.

These low frequency artifacts appear to be due to poor electrode contact with the skin; low frequency artifacts and noise were most prevalent in the participant with significant calf hair (S006). Excess hair on the skin can result in high skin-electrode impedance; additionally, the textile interface is capacitive and the contact improves with increasing frequency [18]. Although the tetrapolar configuration is assumed to eliminate the effect of electrode polarization and skin-electrode impedance above 1kHz [7], poor contact could result in higher impedances that could affect these low frequency measurements.

Because the deviations due to these artifacts are relatively minimal in the magnitude, Cole parameters can be estimated by fitting to the magnitude and discarding low frequency data (see figure 7). While there is no ‘ground truth’ to compare to, the data in figure 6 suggest that if the calf bioimpedance waveform can be faithfully reconstructed by 10 kHz, the Cole parameters can be estimated accordingly.

Artifacts also occurred at high frequencies. In whole body measurements, these artifacts are attributed to a capacitive ‘hook’ effect [28]. However, these
measurements actually show an increase in phase rather than a decrease in phase, more consistent with high frequency artifacts from electrode mismatch [29] and/or cable inductance [30]. Because each electrode was wrapped around the calf, the effective electrode area could be different, contributing to a potential electrode mismatch. Additionally, the skin-electrode impedance could be higher due to uneven contact with the skin due to either roughness of the fabrics or unevenly distributed pressure from the adhesive bands placed around the textiles. As discussed in previous work [12] and similar to the case with low-frequency artifacts, the artifact primarily impacts the magnitude, and a cut-off frequency at e.g. 200 kHz can be used to minimize the impact of these artifacts.

4.6. Study limitations
This study was performed in healthy, young participants without fluid overload or any skin or vascular conditions in the measured areas of the calf. Future studies should test the performance of these textiles in older participants with congestive heart failure and other medical conditions that could impact the performance of the textiles.

5. Conclusion
While conventional Ag/AgCl electrodes provide accurate and reliable bioimpedance measurements, they are not suitable for long term use and do not distribute
current evenly in short body segments. Dry electrodes, such as textile based electrodes, could enable long term bioimpedance measurements for applications such as remote fluid status monitoring. We measured calf bioimpedance in ten healthy volunteers using five different textile materials in a band electrode configuration. We found that band electrodes are preferable over spot electrodes for segmental measurements as the current is more uniformly distributed. We also found that the SMP130T-B fabric performed the best out of the textile materials tested in terms of precision and fit error, but was also the least comfortable and the most expensive. The Stretch fabric had slightly worse performance that the PI30 + B fabric, but was considered much more comfortable and is half as expensive. The SBRM48 fabric had unusable data for 7/10 participants due to a poor connection, possibly due to its low weight. The textiles did have artifacts at low and/or high frequencies for some participants; future research should consider measuring the skin-electrode impedance directly in order to develop improved measurement electronics and/or textile electrodes to reduce or eliminate these artifacts. In the mean time, the impact of these artifacts can be minimized by limiting the frequency range used for fitting the Cole model. Additional studies should confirm the capabilities of these textiles in patients with fluid overload.

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