Original Paper

The Sensitivity of Bioimpedance to Assess Leg Fluid Volume Changes in Children on Hemodialysis: Experimental and Model-Based Investigation

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Introduction. Fluid management plays an essential role in pediatric hemodialysis; however, traditional assessment techniques often lack the precision required for this patient population. Bioimpedance analysis (BIA) offers a non-invasive alternative, but electrode placement significantly impacts measurement sensitivity. This study compared the efficacy of two calf-based electrode configurations — dorsal (ankle) versus plantar (sole) — in detecting fluid volume changes via bioimpedance in pediatric patients, particularly in terms of their sensitivity. Additionally, we evaluated a novel ring electrode arrangement for optimizing sensitivity distribution in lower limb tissues.

Methods. This controlled before-and-after study was conducted at the dialysis unit of Ali Asghar Children's hospital in Tehran, Iran. It evaluated thirteen children (aged 3-15 years) undergoing hemodialysis using bioimpedance analysis. The primary outcome measure was the change in impedance values before and after a single dialysis session. Measurements were performed with electrodes placed in two configurations: between the plantar surface of the foot to below the knee (down), and between the ankle dorsal surface to below the knee (up). Current distribution patterns were modeled using COMSOL simulations.

Results. Mean ultrafiltration volume was 1123 ± 384 mL per session. Impedance changes were observed in both dorsal and plantar electrode configurations, with no significant differences or superiority in measuring these changes. Sensitivity analysis revealed variations of 35% with dorsal placement, 24% with plantar placement, and 20% with ring electrode configuration, demonstrating superior uniformity of the ring design across tissue regions.

Conclusion. Optimal electrode placement varies among patients with no universal advantage for either position. Bioimpedance sensitivity depends on location of fluid accumulation, and enhanced sensitivity is required for monitoring changes below two liters. The ring electrodes provided the most uniform sensitivity distribution. These findings support use of personalized bioimpedance monitoring approaches to improve fluid management precision in pediatric hemodialysis.

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INTRODUCTION

Assessing Body Fluid Changes in Pediatric Hemodialysis Patients

Accurate assessment of body fluid changes in pediatric population on routine hemodialysis (HD) is essential for optimizing ultrafiltration (UF) management. Effective UF adjustments prevent chronic fluid overload complications, including hypertension and cardiovascular diseases, which are prevalent in this population. Monitoring fluid dynamics facilitates optimal fluid balance, improves patient outcomes, and minimizes dialysis-related adverse events.[11] Integrating these measurements into routine practice provides clinicians with an objective, real-time indicator, enhancing precision in fluid management for dialysis care.

Plantar Electrode Configurations and Practical Advantages

Plantar electrode configurations, where electrodes are placed on the soles of the feet, offer practical benefits for bioimpedance monitoring by enabling foot-to-foot impedance measurement. This setup allows patients to simply stand on a device, eliminating the need for gel pads or limb electrodes, which makes it particularly convenient for both clinic and home use.[12] Notably, plantar electrodes can be an effective alternative for repeated measurements in children, as they avoid discomfort often caused by adhesive electrodes in this population, making this method well-suited for pediatric applications. The large surface area of the soles ensures a stable, reproducible interface, enhancing measurement consistency 1300 µA) is applied to the body, it encounters varying degrees of opposition (impedance) depending on tissue composition. Tissues with high water and electrolyte content, such as blood and muscle, offer minimal resistance to electrical flow, while adipose tissue and bone present significantly higher impedance values due to their lower water content.

Most clinical BIA systems operate at a frequency of 50 kHz, which represents an optimal compromise that allows partial penetration of both extracellular and intracellular compartments, thereby enabling a comprehensive assessment of total body water. For accurate measurements, a specific tetrapolar electrode configuration is essential. This setup employs an arrangement, wherein four electrodes are strategically positioned on the skin surface: two for current injection and two for voltage sensing. This configuration minimizes contact impedance errors and enhances measurement accuracy. The impedance (Z) measured in BIA is a complex quantity comprising resistance (R) and reactance (Xc), where $Z^2 = R^2 + Xc^2$.

Resistance represents the opposition to current flow through ionic solutions, while reactance reflects the capacitive properties of cell membranes. At lower frequencies (5-50 kHz), the current primarily flows through extracellular fluid, whereas at higher frequencies (> 100 kHz), it increasingly penetrates cell membranes to include intracellular fluid in the measurement pathway (Figure 1).[4,5]

The evolution of bioimpedance technology has substantially advanced fluid management in dialysis over the past four decades. Initial applications primarily emphasized dry weight determination, whereas the introduction of bioimpedance spectroscopy (BIS) in the early 2000s improved the differentiation between intracellular and extracellular fluid compartments.[6] Recent technological innovations have expanded BIA capabilities to incorporate continuous fluid

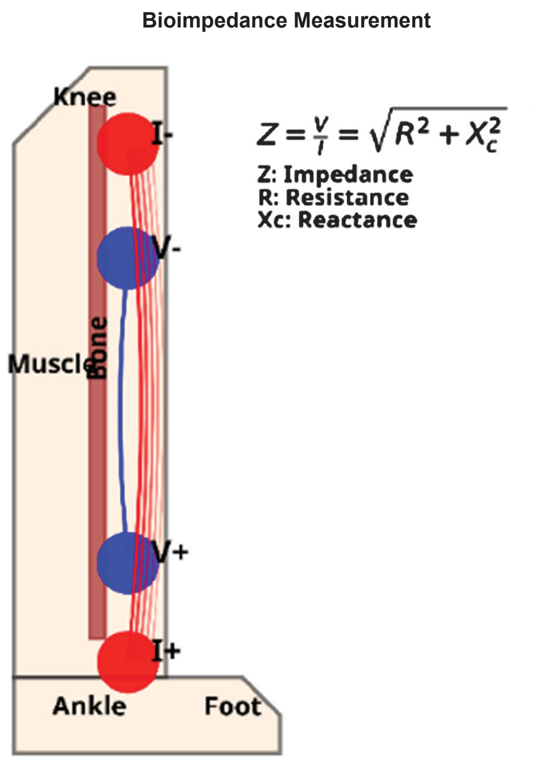

Figure1. Schematic diagram of bioimpedance measurement in the calf region, illustrating electrode placement. Red electrodes (I+, I-) indicate current injection points, and blue electrodes (V+, V-) denote voltage sensing points.

monitoring, with portable devices and digital integration further augmenting clinical accessibility and utility.[7,8]

Calf-Based Bioimpedance for Fluid Volume Assessment

Clinical studies indicate that the calf serves as an ideal site for bioimpedance measurements, particularly for evaluating fluid status in patients on dialysis. This method focuses on changes in the lower leg's tissue resistivity, which strongly correlate with shifts in extracellular fluid volume. For example, Zhu *et al.* showed that continuous calf bioimpedance monitoring during hemodialysis can improve the determination of a patient's dry weight, the target post-dialysis weight indicating optimal hydration.[9] Similarly, Sebert *et al.* observed that normalizing calf impedance readings helps in determining the adequacy of fluid removal.[10] The calf, being one of the last regions to release excess fluid, shows a plateau in bioimpedance that signals stabilized intravascular and interstitial volumes.[11] Integrating these measurements into routine practice provides clinicians with an objective, real-time indicator, enhancing precision in fluid management for dialysis care.

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Pediatric vs. Adult Bioimpedance Considerations

Bioimpedance analysis should consider significant physiological differences between children and adults, as children have a higher proportion of total body water and changing body composition with growth.[15] Standard adult-derived BIA models often misestimate pediatric fluid status due to age-related differences in hydration baselines. Electrode sizing and placement require adjustment for smaller bodies, and keeping young children still during measurement can be challenging. However, BIA is increasingly used in pediatric nephrology as a non-invasive method to guide fluid management, particularly in dialysis, aiding in estimating dry weight and nutritional status with age-appropriate models.[16] Improving these techniques can help tailor ultrafiltration, reduce volume-related complications and bridge care gaps between pediatric and adult dialysis.

Research Objectives

This study focused on children aged 3–15 years receiving regular hemodialysis treatment. Our primary objective was to assess the responsiveness of plantar electrodes to variations in body water compared to standard electrode configurations, including calf-based measurements, and to determine the optimal electrode placement for maximizing measurement sensitivity in this pediatric population. Although plantar electrode placement is commonly employed in bioimpedance devices, its sensitivity to body water changes relative to the standard ankle placement remains underexplored. To address this research gap, we compared these two electrode configurations to establish the most effective positioning for precise fluid assessment. Additionally, we investigated the contributions of various tissues to bioimpedance measurements using anatomically accurate 3D human body models. By prioritizing raw impedance values over derived water volumes, our approach leveraged recent methodological advancements in bioimpedance analysis, aiming to enhance the detection of clinically relevant fluid shifts in this susceptible group. Ultimately, the findings from this study may inform more tailored and accurate fluid management strategies for pediatric hemodialysis patients, potentially improving clinical outcomes and quality of life.

MATERIALS AND METHODS

Study Design and Population

This prospective, cross-sectional, before-after study was conducted at Ali Asghar Children's Hospital, Tehran, Iran. We enrolled 13 children (five males, eight females) aged 3-15 years with end-stage kidney disease (ESKD) undergoing chronic hemodialysis. Inclusion criteria were a stable clinical condition (no hospitalizations or major medication changes for at least one month), regular hemodialysis (≥3 months), and parental consent. Exclusion criteria comprised electronic implants, lower limb abnormalities, foot skin infections, inability to cooperate, or dialysis prescription changes during the study. The study protocol was approved by the Ethics Committee of Iran University of Medical Sciences (IR.IUMS.REC.1402.937).

Study Objectives and Sensitivity Definition

The primary outcome measure was the change in impedance values before and after hemodialysis, compared between dorsal and plantar electrode placements. Secondary outcomes included the spatial distribution of sensitivity in the leg model and the correlation between impedance changes and ultrafiltration volume. "Sensitivity" was interpreted in three specific contexts, consistent with prior bioimpedance literature: (1) The minimum detectable fluid volume change, with a threshold of approximately 1000 mL.[17] (2) The magnitude of impedance change relative to variation in total body water.[18] (3) The influence of tissue properties on impedance measurements. Raw impedance data was used instead of derived parameters because conversion formulas that translate impedance to water measurements may introduce errors, particularly in pediatric ESRD patients for whom standard prediction equations might not be validated.[19]

Impedance Measurement Protocol

Impedance was measured using the MAH12 Bioimpedance Analyzer[20] Iranian Research Organization for Science and Technology (IROST), Tehran, Iran. The device operates at a frequency range of 5-250 kHz with a constant current of 800 µA; and was tested and calibrated at the IROST medical engineering laboratory. Six Ag/AgCl electrodes were placed in 2 configurations (Figure 2): (1) dorsal placement with electrodes on the dorsum of the foot and below the knee(up), and (2) plantar configuration with electrodes on the plantar surface of the foot and below the knee(down).

All measurements were performed with patients in the supine position to standardize body position and minimize movement artifacts, as recommended by Kyle *et al*.[21]

All bioimpedance measurements and numerical simulations were conducted under the supervision of faculty members specializing in bioelectrical engineering with expertise in electrical impedance applications. Their guidance ensured methodological accuracy in both the experimental setup and data interpretation phases of the study.

To account for anthropometric variations in participant, we conducted sensitivity analyses

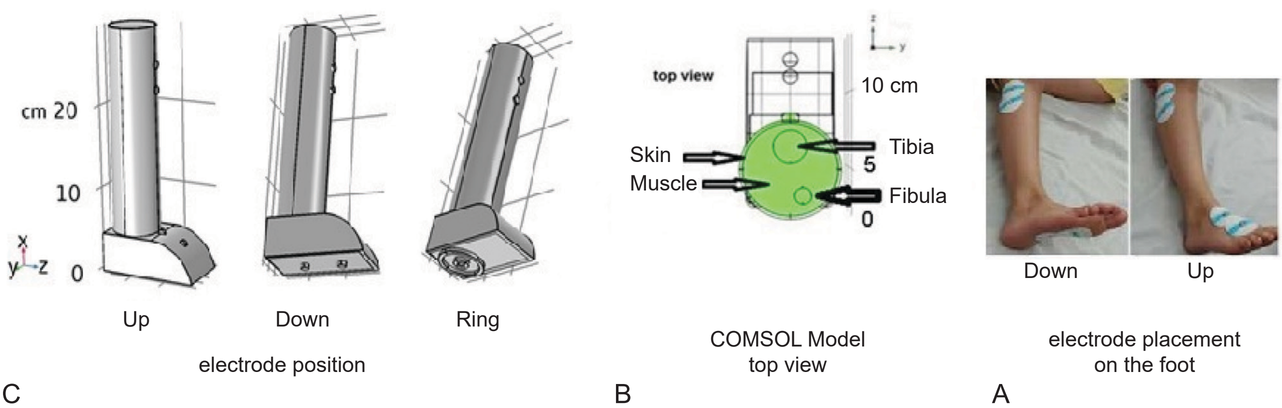

Figure 2. (A) Electrode placement for two configurations - up and down states. (B) Location and dimensions of the COMSOL simulation setup. A cylinder with a radius of 3.1 cm and a length of 25 cm is considered as the lower part of the leg. The outer surface of the leg cylinder is assumed to have a 1 mm space representing the skin layer. Additionally, two cylinders with radii of 1 cm and 0.5 cm are included to represent the tibia and fibula bones within the leg. (C) Electrode shape and location in the COMSOL model.

Table 1. Parameters values in the COMSOL model.

Parameter	Value	Reference
Calf length	25 cm	typ. Children calf length
Foot length	12 cm	typ. Children length
Electrode radius	0.5 cm	approx.size of real electrode
Tibia radius	1 cm	21
Fibula radius	0.5 cm	21
Calf radius	3.1 cm	average from patients
Bone conductivity	0.02 s/m	22
Muscle conductivity,	0.36 s/m	21,4
Skin (wet) conductivity	0.01 s/m	22

with varying model parameters (calf diameter and subcutaneous fat thickness) to reflect the range of characteristics within our study population (Table1).

Our findings do not suggest that manufacturers have incorrect standards, but rather highlight those current standardized approaches may benefit from personalization based on individual patient characteristics. Conventional electrode placement protocols were developed primarily in adult populations whose body compositions differ from pediatric patients with ESKD. Our results suggest that the optimal electrode placement may vary among patients depending on their specific fluid accumulation patterns, which is particularly relevant in pediatric dialysis patients who experience significant growth-related changes.

Numerical simulations were performed using COMSOL Multiphysics (version 5.6) to investigate current distribution and impedance sensitivity under both electrode configurations. The leg model included skin, muscle tissue with anisotropic properties, and bone structures, based on established tissue conductivity values.[22]

$$S = \frac{J_1 . J_2}{I^2}$$

where J_1 and J_2 are current densities and I is the applied current. Resistance was derived by integrating sensitivity over the model volume.

$$R = \int \rho S d\upsilon$$

Simulations assessed sensitivity distribution in longitudinal and transverse sections.

Statistical Analysis

Data were analyzed using Python (version 3.8). All variables were first tested for normality using the Shapiro-Wilk test. For the primary outcome analysis, we compared pre- and post-dialysis impedance values within each electrode configuration (dorsal and plantar) using paired t-tests for normally distributed variables (such as impedance at 50 kHz) and Wilcoxon signed-rank tests for non-normally distributed variables.

To compare the differences between dorsal and plantar electrode placements, we used independent samples t-tests for normally distributed variables and Mann-Whitney U tests for non-normally distributed variables. This approach allowed us to determine which electrode configuration demonstrated greater sensitivity to fluid changes. Statistical significance was set at $P < .05$ for all analyses.

RESULTS

Participant Characteristics

The study included 13 participants (five males and eight females) with mean age 9.3 ± 3.6 years (range:3-15 years). Mean weight was 34.6 ± 11.7 kg and mean height was 115.3 ± 15.6 cm. Cardiovascular parameters were within acceptable limits for this pediatric cohort: mean systolic blood pressure was 115.5 ± 19.2 mmHg, mean diastolic blood pressure was 73.0 ± 21.1 mmHg, and mean heart rate was 90.1 ± 9.3 beats per minute. Dialysis duration was individualized based on clinical requirements, ranging from two to four hours per patient. Hemodialysis sessions removed an average of 1123 ± 384 mL of fluid (range: 400-2000 mL), corresponding to 3.2 ± 0.9% of body weight.

Bioimpedance Measurements

Table 2 summarizes impedance measurements at 50 kHz before and after dialysis for dorsal and plantar configurations. In the dorsal configuration (up), impedance increased significantly post-dialysis (Wilcoxon signed-rank test, $P = .013$), with 10 of 13 patients (76.9%) showing higher values; median change was 4.90 Ω (IQR: 1.50–22.40 Ω). Similarly, In the plantar configuration (down), impedance increased significantly ($P = .002$), with 11 of 13 patients (84.6%) showing increases; median change was 9.00 Ω (IQR: 4.00–13.70 Ω). Comparison of percentage changes between configurations showed no significant difference ($P = .959$), with six patients exhibiting greater changes in the plantar setup and seven in the dorsal setup.

Table 2. Summary of collected |z| data in 50 KHZ. Init: before hemodialysis session, Final: after hemodialysis session.

Patient	UF (lit)	Init up (Ω)	Final up (Ω)	Change (%)	*P*	Init down (Ω)	Final down (Ω)	Change (%)	*P*
1	1.1	211 ± 0.61	208 ± 0.25	-1.5 ± 0.17	**	220.4 ± 0.04	223.5 ± 0.04	1.3 ± 0.02	**
2	0.8	358.1 ± 2.23	408.9 ± 1.8	14.2 ± 1	**	328.5 ± 1.27	378.6 ± 1.36	15.2 ± 0.61	**
3	1	149.2 ± 0.11	151.5 ± 0.0	1.5 ± 0.07	**	156.3 ± 0.06	160.9 ± 0.04	2.9 ± 0.06	**
4	1	197.3 ± 0.1	201.2 ± 0.07	1.9 ± 0.08	**	233.3 ± 0.07	231.2 ± 0.06	-0.9 ± 0.04	**
5	0.8	260 ± 0.1	282.4 ± 0.16	8.5 ± 0.03	**	260 ± 0.09	264 ± 0.24	1.3 ± 0.06	**
6	1.5	209 ± 0.1	213.9 ± 0.1	1.9 ± 0.02	**	192.8 ± 0.12	205.4 ± 0.2	6.5 ± 0.22	**
7	1.4	217.3 ± 0.1	271.6 ± 0.35	24.9 ± 0.1	**	282.8 ± 0.08	291.8 ± 0.2	3.21 ± 0.09	**
8	1.0	200.2 ± 0.04	205.8 ± 0.12	2.76 ± 0.05	**	358.7 ± 0.23	367.1 ± 0.07	2.35 ± 0.05	**
9	2	43.3 ± 0.1	40.4 ± 0.1	-7.6 ± 0.03	**	46.4 ± 0.1	60.1 ± 0.12	29.3 ± 0.7	**
10	1.1	127.9 ± 0.1	150.5 ± 0.12	17.6 ± 0.8	**	137 ± 0.14	157 ± 0.16	14.5 ± 0.4	**
11	1.5	78.7 ± 0.12	80.2 ± 0.13	1.9 ± 0.03	**	146 ± 0.11	141.4 ± 0.15	-3.15 ± 0.1	**
12	0.4	137.5 ± 0.2	144.3 ± 0.2	4.9 ± 0.04	**	294.1 ± 0.3	306.7 ± 0.2	4.28 ± 0.4	**
13	1	85.3 ± 0.1	83.1 ± 0.2	-2.5 ± 0.03	**	191 ± 0.3	208 ± 0.2	8.9 ± 0.1	**
Mean ± SD	1.12 ± 0.38	174.9 ± 80.8	187.8 ± 94.1	5.4 ± 8.4	0.013	219.0 ± 83.3	230.4 ± 87.5	6.5 ± 8.3	.002

** = $P < 0.001$ - Note: P-values from Wilcoxon signed-rank test; dorsal vs. plantar percentage change $P = 0.959$.

Despite the overall trend of increasing impedance, three patients with dorsal placement and two patients with plantar design showed an unexpected decrease in impedance after dialysis. This suggests potential influences beyond fluid removal, such as fluid redistribution or tissue conductivity changes (Figure 3).

Simulation Results

To better understand the physical basis for the observed bioimpedance patterns, we conducted detailed numerical simulations of current distribution and sensitivity in the lower limbs.

Longitudinal Plane Analysis

Numerical simulations in the Longitudinal Plane (Figure 4) demonstrated that bioimpedance measurements exhibit heightened sensitivity to fluid changes in tissues located near the electrodes or in curved body regions, likely due to higher current density in these areas. The simulation revealed up to a tenfold difference in impedance sensitivity within the leg tissues, with values ranging from $10^6 \frac{1}{m^4}$ near the electrode to $10^5 \frac{1}{m^4}$ in tissues farther away.

Transverse Plane Analysis

Analysis of the Transverse Plane (Figure 5)

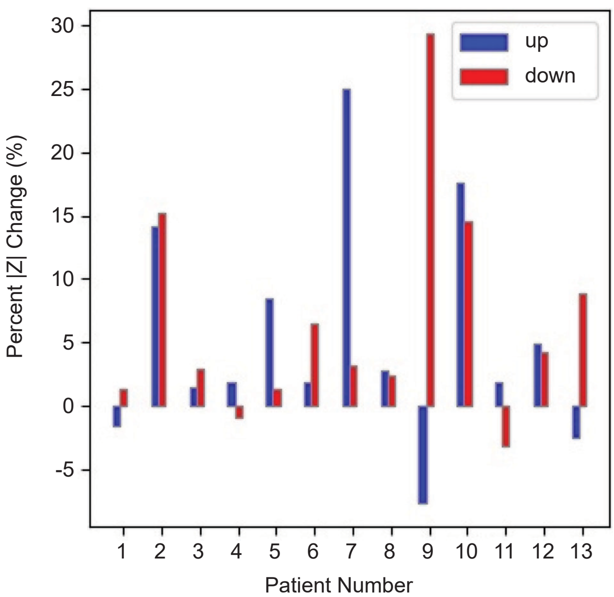

Figure 3. Percentage change in impedance (|Z|) at 50 kHz, comparing dorsal (up) and plantar (down) electrode placements. Seven patients showed greater changes with dorsal placement, while six exhibited larger changes with plantar placement.

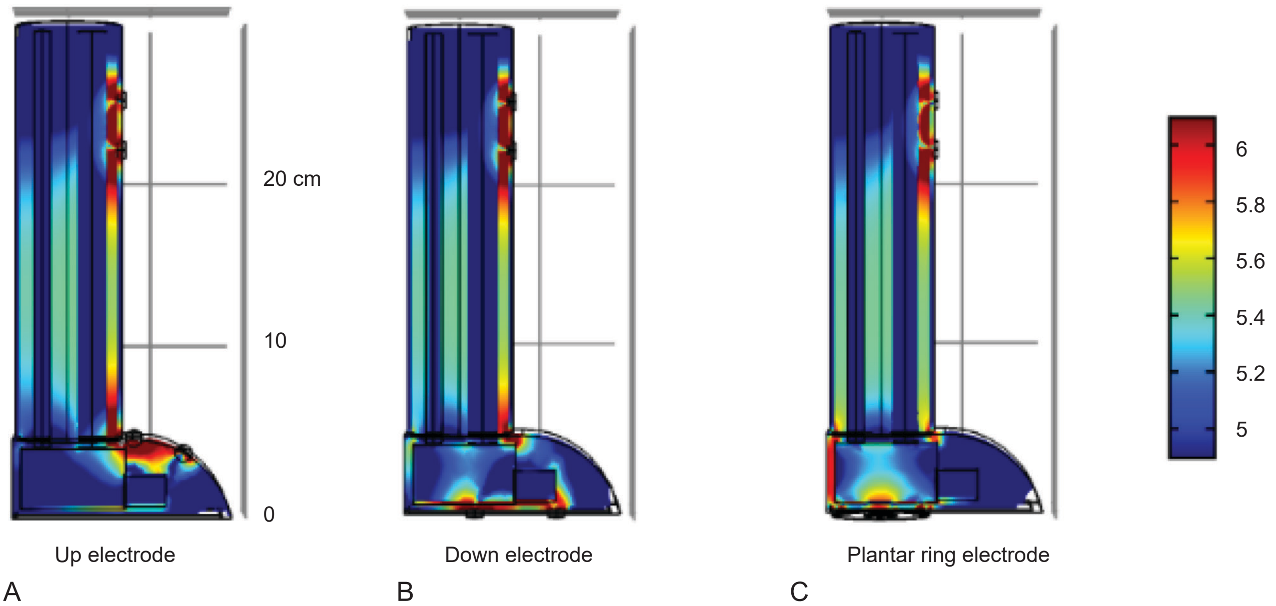

Figure 4. Impedance sensitivity distribution (logarithmic scale) in the longitudinal plane: (A) dorsal-up, (B) plantar-down, (C) ring electrode configurations. Sensitivity is non-uniform in lower leg regions, with higher values near electrodes and ankle curvature; ring electrodes improve uniformity.

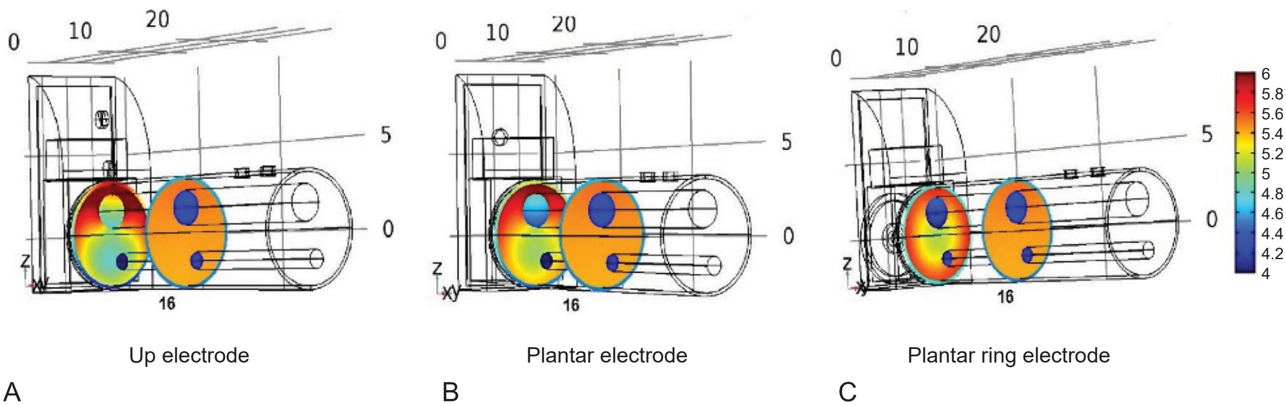

Figure 5. Impedance sensitivity (log scale) in the transverse plane: Uniformity emerges ~16 cm from the sole, with ring electrodes enhancing consistency in lower leg muscle tissues.

revealed that beyond a distance of 16 cm from the plantar surface of the feet, the sensitivity distribution in the muscle section of the lower limbs became uniform, regardless of electrode location. This transition point represents an important boundary for electrode placement considerations in patients.

Sensitivity Analysis in the Calf Region

Given that the calf region is a common site for fluid accumulation in hemodialysis patients, we conducted a specific sensitivity analysis using a hypothetical cross-section in this region (Figure 6A). The analysis revealed that the impedance sensitivity in the calf were 35% in the dorsal configuration, 24% in the plantar configuration, and 20% in the ring electrode configuration. This quantitative comparison demonstrates that the ring electrode configuration, in which the electrodes are positioned on the plantar surface of the foot in a circular arrangement, provides the most uniform sensitivity distribution.

DISCUSSION

Methodological Considerations and Key Findings

This study demonstrated that bioimpedance analysis (BIA) effectively detects fluid alterations in

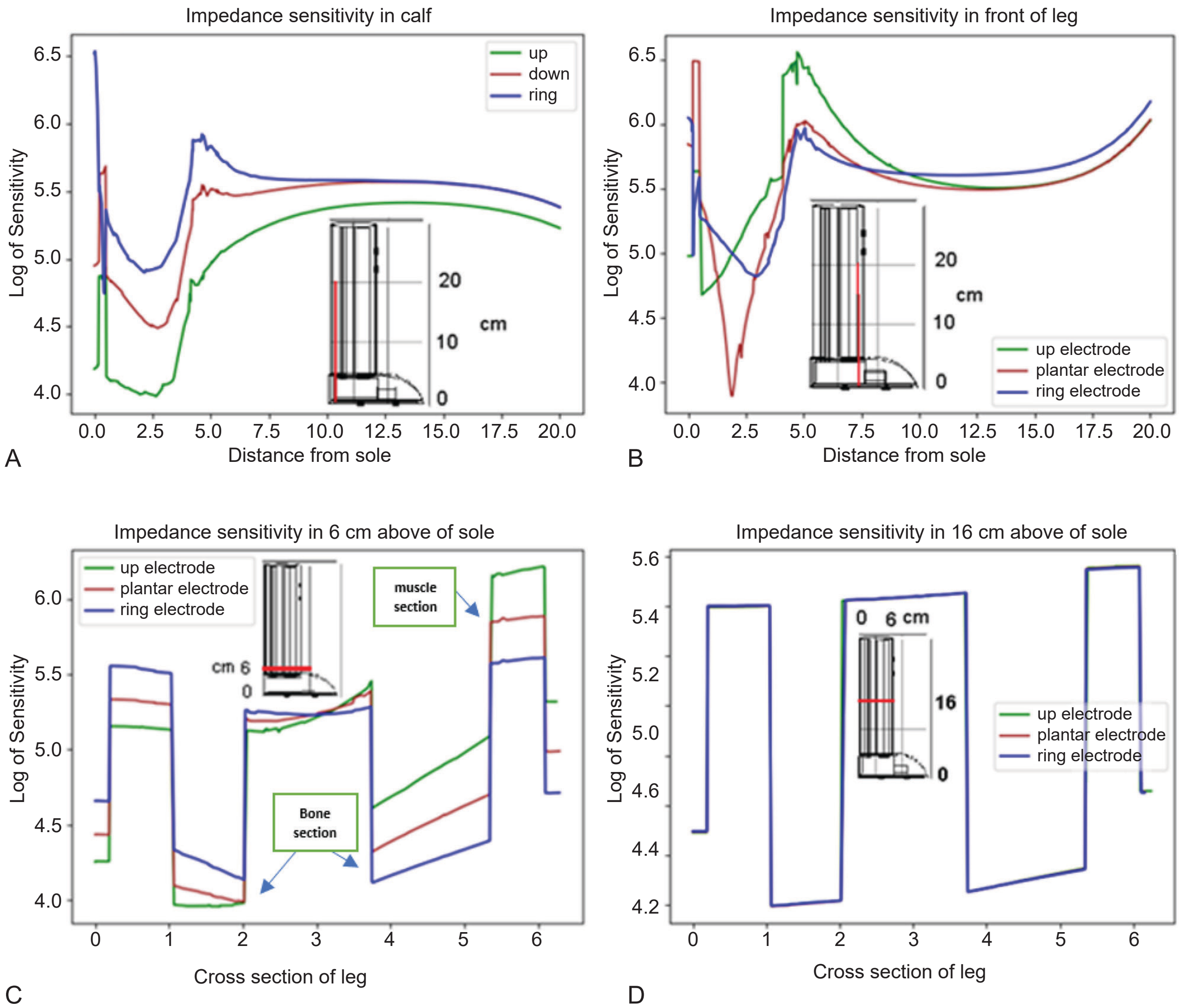

Figure 6. Sensitivity along a cut line (red) in the leg's longitudinal and transverse sections. (A) Calf; (B) Lower and anterior foot; (C) six cm above the sole; (D) 16 cm above the sole. (A, B) Dorsal and plantar electrodes show higher sensitivity near the ankle (see Figure 5A, B). (C) At six cm, dorsal setup shows 10-unit sensitivity differences due to tissue non-uniformity (see Figure 5C). (D) At 16 cm, sensitivity is more uniform in the plantar setup (see Figure 5D).

pediatric patients undergoing regular hemodialysis, as evidenced by significant post-dialysis increases in impedance using both dorsal (median change: 4.90 Ω, 5.4%, $P = .013$) and plantar (median change: 9.00 Ω, 6.5%, $P = .002$) electrode configurations. Due to the non-normal distribution of data (Shapiro-Wilk test, $P < .05$), we employed the Wilcoxon signed-rank test, which uncovered significant changes that might have been overlooked by parametric tests.[23]

These group-level findings are consistent with prior research reporting increased impedance post-dialysis due to fluid removal.[24,25] Notably, Schotman *et al.* also confirmed that a 50 kHz frequency, as used in our study, provides sufficient sensitivity for detecting extracellular water (ECW) changes in hemodialysis patients, supporting the reliability of our frequency selection.[26]

Although most patients exhibited the expected impedance increase, individual variability was evident as three participants in the dorsal and two in the plantar configurations showed decreases. This may be attributable to the dialysate sodium concentration (143 mEq/L) used to prevent ultrafiltration-induced hypotension. Elevated sodium concentrations in the extracellular compartment (resulting from the high dialysate sodium concentration) can enhance extracellular conductivity or cause osmotically driven fluid redistribution between intracellular and extracellular compartments[27], reducing the impedance and potentially leading to misinterpretation of fluid status.[28,29]

A methodological strength of the current study lies in its focus on relative impedance changes rather than absolute values, an approach also adopted by Schotman *et al.*, who similarly analyzed raw impedance data to avoid errors from conversion formulas.[26] In contrast to many studies that estimate dry weight or total body water using algorithms derived from healthy populations, which often yield inaccurate results for dialysis patients due to altered fluid dynamics (Mitsides *et al.*)[30] we assessed percentage changes in impedance within each patient, using individuals as their own reference. This method minimizes the impact of body size and tissue variability, calibrating impedance changes against individual ultrafiltration volumes[31] By evaluating the percentage variation in impedance for each patient relative to their fluid removal, we can monitor these changes across consecutive sessions to develop a model of the patient's body water dynamics over time. When tracking these relative changes over sessions, our approach provides the groundwork for modeling a patient's body water dynamics over time, offering a more personalized and dynamic assessment of fluid shifts during dialysis.[32]

Our approach does not challenge manufacturer standards for absolute bioimpedance measurements but rather proposes a complementary method focused on dynamic changes in individual patients, which may be more clinically relevant for monitoring fluid shifts during dialysis.

Electrode Configuration and Clinical Implications

Our findings regarding electrode placement offer significant clinical insights for pediatric hemodialysis. The lack of a significant difference between dorsal and plantar configurations (P = .959) suggests flexibility in electrode placement, enhancing measurement feasibility in clinical settings, particularly for children where ease of electrode application is crucial.

Our research on electrode placement provides substantial clinical insights for pediatric hemodialysis. The lack of a significant difference between dorsal and plantar configurations (P = .959) suggests flexibility in electrode placement, enhancing measurement feasibility in clinical settings, particularly for children in which the ease of electrode application is important. Through numerical simulations, we quantitatively assessed how fluid-induced tissue changes affect impedance measurements, revealing that electrode positioning alters the projection of these changes in impedance values. Specifically, the simulations (Figures 4 and 5) showed that tissues in curved regions, such as the ankle, exhibit heightened sensitivity—up to tenfold higher than surrounding tissues—due to higher current density, a finding consistent with limited prior studies.[33] Areas of high sensitivity are highlighted in red in Figure 4, indicating that electrodes placed on the dorsum of the foot better reflect fluid variations in the forefoot, while plantar electrodes are more sensitive to changes in the hindfoot (Figure 4).

A critical aspect of this study is the uniformity of sensitivity, which is essential for accurate measurement of fluid alteration across different body regions. While segmental BIA has improved sensitivity uniformity to some extent, it still lacks the precision required to detect small fluid change. Our simulations demonstrate that non-uniformity persists within segments such as the leg, particularly near the electrodes. However, plantar electrodes, due to their alignment with the leg's longitudinal axis, achieve uniform sensitivity at a shorter distance from the sole (Figures 6A, 6B). Beyond approximately 16 cm from the sole, sensitivity becomes uniform across all three configurations (dorsal, plantar, and ring), with no notable differences (Figure 6D). Thus, the primary variability in impedance measurements arises from regions closest to the electrodes (Figure 6D). Notably, the ring electrode configuration provided the most uniform sensitivity distribution (20% variation) compared to dorsal (35%) and plantar (24%) setups, with sensitivity gradients persisting up to 16 cm above the sole. This aligns with recent studies, such as Usman *et al.*[8,34]), who reported a strong correlation (r = 0.90) between a ring bioimpedance analyzer and reference devices for body fat estimation, and Sel *et al.* who demonstrated a correlation of 0.81 for ring-shaped sensors in blood pressure monitoring.[35] The ring configuration in our study, featuring two concentric hollow rings, ensures a more balanced current distribution, reducing dependency on specific anatomical fluid patterns and enhancing measurement reliability.

Limitations and Future Directions

Despite its contributions, this study has limitations that warrant consideration. The small sample size (n = 13), while reflective of our center's pediatric hemodialysis cohort, limits the generalizability the results of our study. Additionally, the simulation model assumed uniform tissue properties, potentially oversimplifying the anatomical complexity and tissue anisotropy in children. Variability in dialysate composition, particularly sodium levels, was not fully adjusted for and might have affected impedance measurements, as observed in patients with impedance decreases. Future research should focus on: (1) larger, longitudinal studies incorporating diverse electrode configurations and multi-frequency BIA to enhance sensitivity; (2) development of personalized algorithms to model individual fluid patterns over time; and (3) exploration of advanced electrode designs, such as wearable or hybrid configurations, to minimize variability in pediatric populations .Our findings underscore the need for individualized BIA approaches, paving the way for refined fluid management strategies in pediatric dialysis care.

CONCLUSION

This study revealed that the impact of electrode placement (dorsal, plantar, or ring) on bioimpedance analysis varies among pediatric hemodialysis patients, depending on the anatomical location of fluid accumulation. Impedance sensitivity became uniform across all configurations beyond 16 cm from the sole of the foot, with the ring electrodes on the plantar surface providing the most consistent sensitivity distribution. Experimental and simulation results further demonstrated that impedance changes are influenced not only by the volume of the removed fluid but also by the site of fluid accumulation and alterations in extracellular conductivity. These findings highlight the importance of considering individual patient characteristics and fluid distribution patterns when selecting electrode placements for impedance-based fluid assessments, emphasizing the need for personalized approaches to optimize fluid management in hemodialysis care.

REFERENCES

1. Fischbach M, Zaloszyc A, Shroff R. The interdialytic weight gain: a simple marker of left ventricular hypertrophy in children on chronic haemodialysis. doi:10.1007/s00467
2. Jia M. The relationship of volume overload and its control to hypertension in hemodialysis patients. Physiol Behav. 2017;176(3):139-148. doi:10.1111/sdi.12838.The
3. Jaeger JQ, Mehta RL. Assessment of dry weight in hemodialysis: an overview. Journal of the American Society of Nephrology. 1999;10(2):392-403.
4. Simini F, Bertemes-Filho P. Bioimpedance in Biomedical Applications and Research.; 2018. doi:10.1007/978-3-319-74388-2
5. Seoane F, Abtahi S, Abtahi F, et al. Mean expected error in prediction of total body water: A true accuracy comparison between bioimpedance spectroscopy and single frequency regression equations. Biomed Res Int. 2015;2015. doi:10.1155/2015/656323
6. Davies SJ, Davenport A. The role of bioimpedance and biomarkers in helping to aid clinical decision-making of volume assessments in dialysis patients. Kidney Int. 2014;86:489-496. doi:10.1038/ki.2014.207
7. Chauhan R. Wearable Embedded System Design and Development for Hydration Monitoring via Bio impedance Analysis. Turkish Journal of Computer and Mathematics Education (TURCOMAT). 2018;9(3):1032-1039. doi:10.17762/turcomat.v9i3.13891
8. Usman M, Thapa S, Gupta AK, Xue W. Ring Based Wearable Bioelectrical Impedance Analyzer for Body Fat Estimation. In: 2018 IEEE International Symposium on Signal Processing and Information Technology (ISSPIT). 2018:291-296. doi:10.1109/ISSPIT.2018.8642675
9. Zhu F, Levin NW. Estimation of body composition and normal fluid status using a calf bioimpedance technique. In: Blood Purification. Vol 39. S. Karger AG; 2015:25-31. doi:10.1159/000368937
10. Seibert E, Müller SG, Fries P, et al. Calf bioimpedance spectroscopy for determination of dry weight in hemodialysis patients: Effects on hypertension and left ventricular hypertrophy. Kidney Blood Press Res. 2013;37(1):58-67. doi:10.1159/000343400
11. Seibert E, Zhu F, Kuhlmann MK, et al. Slope analysis of blood volume and calf bioimpedance monitoring in hemodialysis patients. Nephrology Dialysis Transplantation. 2012;27(12):4430-4436. doi:10.1093/ndt/gfr734
12. Diaz DH, Casas O, Pallas-Areny R. Heart rate detection from single-foot plantar bioimpedance measurements in a weighing scale. Annu Int Conf IEEE Eng Med Biol Soc. 2010;2010:6489-6492. doi:10.1109/IEMBS.2010.5627358
13. Jaffrin M, Bousbiat S. Accuracy of plantar electrodes compared with hand and foot electrodes in fat-free-mass measurement. J Healthc Eng. 2014;5(2):123-144. doi:10.1260/2040-2295.5.2.123
14. Iacopino L, Andreoli A, Innocente I, et al. Use of foot-to-foot bioelectrical impedance analysis in children. In: Acta Diabetologica. Vol 40. 2003. doi:10.1007/s00592-003-0068-0
15. Orsso CE, Gonzalez MC, Maisch MJ, Haqq AM, Prado CM. Using bioelectrical impedance analysis in children

and adolescents: Pressing issues. Eur J Clin Nutr. 2022;76(5):659-665. doi:10.1038/s41430-021-01018-w

16. Gutiérrez-Marín D, Escribano J, Closa-Monasterolo R, et al. Validation of bioelectrical impedance analysis for body composition assessment in children with obesity aged 8-14y. Clinical Nutrition. 2021;40(6):4132-4139. doi:https://doi.org/10.1016/j.clnu.2021.02.001
17. Schotman JM, Borren MMGJ, Wetzels JFM, Kloke HJ, Reichert LJM, Boer H. Sensitivity of total body electrical resistance measurements in detecting extracellular volume expansion induced by infusion of NaCl 0 . 9 % Sensitivity of total body electrical resistance measurements in detecting extracellular volume expansion induced b. Eur J Clin Nutr. 2020;(August). doi:10.1038/s41430-020-0655-y
18. Bradbury MG, Smye SW, Brocklebank JT. Assessment of the sensitivity of bioimpedance to volume changes in body water. Pediatric Nephrology. 1995;9(3):337-340. doi:10.1007/BF02254204
19. Mobarak M, Researcher P. A Review Paper on the Sensitivity Analysis in Bioimpedance Measurement Technique. Vol 7.; 2022. doi:10.5281/zenodo.6604899
20. Nori, Shahrokh(IROST BED. MAH12 Bioimpedance Analyzer.; 2023.
21. Preedy VR. Handbook of Anthropometry: Physical Measures of Human Form in Health and Disease. Springer New York, NY; 2012. doi:10.1007/978-1-4419-1788-1
22. Gabriel S, Lau RW, Gabriel C. The dielectric properties of biological tissues: III. Parametric models for the dielectric spectrum of tissues. Phys Med Biol. 1996;41(11):2271-2293. doi:10.1088/0031-9155/41/11/003
23. Kyle UG, Bosaeus I, De Lorenzo AD, et al. Bioelectrical impedance analysis - Part I: Review of principles and methods. Clinical Nutrition. 2004;23(5):1226-1243. doi:10.1016/j.clnu.2004.06.004
24. Rutkove S, Pacheck A, Sanchez B. Sensitivity distribution simulations of surface electrode configurations for electrical impedance myography HHS Public Access. doi:10.1002/mus.25561
25. Field AP. Discovering Statistics Using SPSS (and Sex and Drugs and Rock “n” Roll). SAGE; 2012.
26. Moissl U, Arias-Guillén M, Wabel P, et al. Bioimpedance-guided fluid management in hemodialysis patients. Clinical Journal of the American Society of Nephrology. 2013;8(9):1575-1582. doi:10.2215/CJN.12411212
27. Paglialonga F, Ardissino G, Galli MA, Scarfia R V., Testa S, Edefonti A. Bioimpedance analysis and cardiovascular status in pediatric patients on chronic hemodialysis. Hemodialysis International. 2012;16(SUPPL. 1). doi:10.1111/j.1542-4758.2012.00743.x
28. Schotman JM, Hazeleger LR, van Borren MMGJ, et al. Optimal current frequency for the detection of changes in extracellular water in patients on hemodialysis by measurement of total body electrical resistance. Clin Nutr ESPEN. 2021;43:302-307. doi:10.1016/j.clnesp.2021.03.035
29. Ozturk S, Taymez DG, Bahat G, et al. The influence of low dialysate sodium and glucose concentration on volume distributions in body compartments after haemodialysis: a bioimpedance analysis study. Nephrol Dial Transplant. 2008;23(11):3629-3634. doi:10.1093/ndt/gfn274
30. Mitsides N, McHugh D, Swiecicka A, et al. Extracellular resistance is sensitive to tissue sodium status; implications for bioimpedance-derived fluid volume parameters in chronic kidney disease. J Nephrol. 2020;33(1):119-127. doi:10.1007/s40620-019-00620-3
31. Sarkar S, Wystrychowski G, Zhu F, Usvyat L, Kotanko P, Levin N. Fluid Dynamics During Hemodialysis in Relationship to Sodium Gradient Between Dialysate and Plasma. ASAIO Journal. 2007;53:339-342. doi:10.1097/MAT.0b013e318033cba7
32. Mitsides N, Mchugh D, Swiecicka A, Mitra R, Brenchley P. Extracellular resistance is sensitive to tissue sodium status ; implications for bioimpedance derived fluid volume parameters in chronic kidney disease. J Nephrol. 2020;33(1):119-127. doi:10.1007/s40620-019-00620-3
33. Novak I, Davies PSW, Elliott MJ. Noninvasive estimation of total body water in critically ill children after cardiac operations: Validation of a bioelectric impedance method. J Thorac Cardiovasc Surg. 1992;104(3):585-589. doi:https://doi.org/10.1016/S0022-5223(19)34722-1
34. Cha K, Chertow G, Gonzalez J, Lazarus J, Wilmore D. Multifrequency bioelectrical impedance estimates the distribution of body water. J Appl Physiol. 1995;79 4:1316-1319. doi:10.1152/JAPPL.1995.79.4.1316
35. Rutkove SB, Pacheck A, Sanchez B. Sensitivity distribution simulations of surface electrode configurations for electrical impedance myography. Muscle Nerve. 2017;56(5):887-895. doi:10.1002/mus.25561
36. Sel K, Osman D, Huerta N, Edgar A, Pettigrew RI, Jafari R. Continuous cuffless blood pressure monitoring with a wearable ring bioimpedance device. NPJ Digit Med. 2023;6(1). doi:10.1038/s41746-023-00796-w
37. Osman D, Jankovic M, Sel K, Pettigrew R, Jafari R. Blood Pressure Estimation using a Single Channel Bio-Impedance Ring Sensor. 2022 44th Annual International Conference of the IEEE Engineering in Medicine & Biology Society (EMBC). Published online 2022:4286-4290. doi:10.1109/EMBC48229.2022.9871653

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