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Effects of a passive neck support exoskeleton on neck-shoulder muscle fatigue and motion during live veterinary surgeries

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1. Introduction

Musculoskeletal symptoms (MSS) are widely prevalent among veterinary surgeons, with reported rates ranging from 49 to 93% (Jones et al., 2020; Kim et al., 2023; Scuffham et al., 2010; Seagren et al., 2022). Physical health risks, particularly MSS, can adversely impact the surgical performance and well-being of veterinary surgeons (Wallace, 2019). Arias et al. (2023) reported that MSS were associated with stress and chronic pain affecting multiple body regions in all surveyed participants. MSS also reduced professional efficiency and performance, leading to prolonged work absences, sometimes lasting up to 12 months, and difficulties in performing surgical procedures. Beyond functional impairment, MSS imposed additional clinical and financial burdens, with 48.6% of veterinary surgeons reporting the use of medications and 34% undergoing physical therapy to alleviate symptoms (Kim et al., 2023).

Physical workload is a major risk factor for MSS in surgical work on both veterinary and human patients (Gorce and Jacquier-Bret, 2025; Thielmann et al., 2024). While MSS have been extensively investigated in surgical environments for human patients (Yu et al., 2016a; Yu et al., 2016b; Yu et al., 2017; Alleblas et al., 2017; Epstein et al., 2018; Jacquier-Bret and Gorce, 2023), the diversity of surgical settings for veterinary patients further complicates ergonomic safety and makes it more challenging to design and implement effective interventions (Reijula et al., 2003; Villarreal et al., 2023). Many studies have identified risk factors contributing to MSS among veterinary surgeons, including awkward postures, repetitive tasks (Randall et al., 2012), and prolonged static procedures (Asadi et al., 2022). Among these physical risk factors for veterinary surgeons, sustained static neck flexion, namely, leaning the head forward to view the surgical site, is a common contributor to neck MSS (Zheng et al., 2024). Veterinary surgeons are often constrained to forward head postures, which were shown to increase musculoskeletal load in the spinal regions for minimally invasive surgeons (Van Veelen et al., 2003). Asadi et al. (2022) found a mean of 17 degrees of neck flexion in veterinary surgeries, which is close to the 20-degree threshold of high-risk neck postures (McAtamney and Corlett, 1993). Research indicates that maintaining a forward head posture for extended periods can lead to muscle fatigue and alterations in cervical spine curvature, contributing to neck pain (Zhou et al., 2024). Ergonomic interventions against sustained forward head postures are needed for reducing veterinary surgeons' neck MSS.

Ergonomic interventions have been proposed to reduce neck-related musculoskeletal risks in veterinary surgery through both equipment- and behavior-based strategies. In veterinary dentistry, sustained neck flexion is a major concern, and the introduction of ergonomically designed seating, such as saddle-sitting or sit-stand chairs, helps preserve spinal curvature and reduce cervical strain (DeForge, 2002). Similarly, magnification systems with adjustable declination angles and coaxial illumination allow clinicians to maintain a neutral gaze without excessive neck bending (DeForge, 2002). In large-animal field surgeries, posture-supporting devices such as ergonomic stools, foreskin support blocks, immobilization aids, and protective pads have been developed to reduce cervical and lumbar flexion (Vulcani et al., 2016). A recent study demonstrated that integrating short active microbreaks (including stretching, strengthening, and relaxation exercises) along with weekly ergonomics education significantly decreased the prevalence of neck and back pain among veterinary students while enhancing their self-efficacy in handling physically demanding and risky clinical situations (Grünwald and Franziska Licka, 2023). However, these interventions have notable limitations. Ergonomic tools frequently lack flexibility, making them unsuitable for the diverse range of surgical procedures encountered in veterinary practice. Furthermore, integrating permanent ergonomic solutions typically requires significant financial investment, which may not be feasible for all veterinary facilities (Sancibrian et al., 2014; Stone and McCloy, 2004). Additionally, existing studies provide limited empirical evidence demonstrating the effectiveness of traditional ergonomic interventions in veterinary surgery contexts (Cha et al., 2020).

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Exoskeletons offer a promising alternative to traditional ergonomic interventions by providing adjustable, portable, and adaptable support suitable for diverse surgical environments. Prior research has demonstrated their biomechanical potential to reduce muscle activation and fatigue during surgical tasks in the arm and shoulder regions in human surgery. In live surgical settings, a passive arm-support exoskeleton has been shown to reduce elevated-arm exposure and peak muscle activity in the trapezius and deltoid muscles while maintaining usability (Cha et al., 2024). In simulated vascular surgery tasks, Tetteh et al. (2022) demonstrated that neck, arm, and trunk exoskeletons differentially reduced muscle-activation demands, with the trunk device significantly decreasing EMG fatigue rates, indicating the biomechanical potential of passive supports under controlled conditions. A recent study also found that exoskeletons can help reduce surgeons' neck discomfort and improve neck and torso postures while operating on human patients (Kim et al., 2025). However, the effects of neck support exoskeletons on muscle activation and fatigue in surgeons during surgical procedures involving either human or animal patients have not been quantified. Comprehensive evaluations that integrate subjective responses, muscular, and postural factors are needed as they can provide a more robust assessment in real-world surgical settings. Furthermore, no studies have assessed the impact in actual veterinary workplaces, where musculoskeletal injury risks may be further enhanced due to the unique tasks and environmental constraints when operating on animal patients.

Overall, exoskeletons show great promise thanks to their adaptability and biomechanical advantages, as they may reduce muscle activity levels, minimize fatigue, and potentially improve posture. However, important research gaps persist. This study aimed to evaluate the effectiveness of a passive exoskeleton in providing neck support. We hypothesized that (1) using a passive neck-support exoskeleton would reduce self-reported discomfort, as well as muscle activation and fatigue in the neck-shoulder region, without restricting the neck joint range of motion; (2) the exoskeleton does not increase the muscle activity in the lower back.

2. Methods

2.1. Participants

This study was approved by the Institutional Review Board (IRB #1807587226). Informed consent was obtained from all participants prior to each surgical procedure. A total of 36 surgical procedures, including 16 orthopedic and 20 soft-tissue procedures, were performed by ten veterinary surgeons (seven males, three females; four attending surgeons and six resident surgeons) from Purdue University College of Veterinary Medicine. Their demographic information is summarized in Table 1.

2.2. Study design

A within-subject, crossover study design was used to test the hypotheses. Each participant completed surgical procedures under two experimental conditions: (1) a control condition, in which participants wore sensors but did not use the exoskeleton; and (2) an exoskeleton intervention condition, in which participants wore both sensors and the exoskeleton during surgery. The sequence of conditions was randomized for each participant to minimize order effects. Each surgeon completed between two and four surgical procedures, depending on the availability and nature of procedures encountered during the study period.

Table 1
Demographic information of ten veterinary surgeons.

Characteristics	Value
Age, mean (SD), yrs	39.5 (13.6)
Height, mean (SD), meters	1.8 (0.1)
Weight, mean (SD), kilograms	79.6 (16.4)
Years of experience, mean (SD)	10.7 (13.7)

2.3. Study procedures

Upon arrival, participants were briefed on the experimental protocol, after which written informed consent was obtained. Participants then completed a pre-surgical questionnaire adopted from Trejo et al. (2006) and Beurskens et al. (2000). The questionnaire included musculoskeletal symptoms and was administered before and after each surgical procedure (pre- and post-surgery) and was applied to previous surgical ergonomic studies (Doné et al., 2004; Asadi et al., 2022). Participants rated the intensity of discomfort or symptoms for each body region and domain using three ordinal categories: "No," "Slight," or "Substantial." For the upper extremities, ratings were recorded separately for the left (L) and right (R) sides for the following items: shoulder/arm stiffness and shoulder/arm pain. For the neck and trunk, single ratings were provided for neck stiffness, neck pain, back stiffness, and back pain (Appendix A2). Ratings were later coded numerically (0 = No, 1 = Slight, 2 = Substantial) to compute within-subject differences between pre- and post-surgery conditions. Subsequently, wearable surface electromyography (EMG) sensors and inertial measurement units (IMUs) were placed on participants, followed by a standardized sensor calibration procedure (detailed in the Data Collection section). For trials involving exoskeleton intervention, participants were fitted with the passive exoskeleton (Levitate Tech Inc., Fig. 1), with necessary adjustments made by research team members. Following preparation, participants scrubbed and performed the surgical procedures. After the surgery, a post-surgical questionnaire was administered to assess musculoskeletal discomfort in the same anatomical regions, as well as to

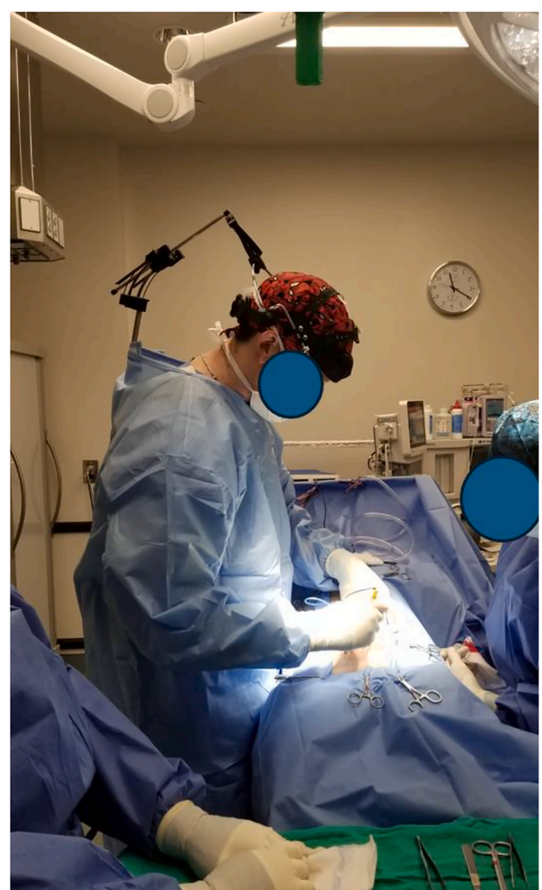

Fig. 1. One of the surgeons is performing surgery wearing the exoskeleton.

collect demographic information. Finally, the wearable sensors and, if applicable, the exoskeleton were removed from the participant.

2.4. Data collection

To capture biomechanical and physiological data during surgery, participants were equipped with a combination of surface electromyography (EMG) sensors (1000 Hz) and inertial measurement units (IMUs) (100–120 Hz). DataLITE Wireless EMG Sensors (Biometrics Ltd., Newport, UK) were applied to assess muscle activity, including maximum voluntary electrical activation (MVE) and median power frequency (MPF), in the neck-shoulder and back. Electrodes were positioned in accordance with SENIAM recommendations: Specifically, two EMG sensors were placed at the midpoint of the line connecting the acromion and the seventh cervical vertebra (C7) (Mercer, 2002). They were oriented following the muscle fibers and tilted about 10–20° upward, slightly inward (Fig. 2). The other two EMG sensors were positioned bilaterally on the lumbar paraspinals, at approximately 2–3 cm lateral to the spinous process of the L2 vertebra (Hermens et al., 1999). EMG sensors were aligned parallel to the muscle fiber orientation and mounted on the skin using double-sided adhesive pads with cut-out holes allowing direct electrode-skin contact.

Due to sensor availability, two different IMU systems were used: the Xsens Awinda system (Movella Inc., El Segundo, USA) and the APDM Opal system (Clario Inc., Philadelphia, USA). Four participants were equipped with the Xsens system, which included seven wireless IMUs configured for full upper-body kinematic tracking (excluding the hands and forearms). Sensors were placed on the head, sternum, shoulders, upper arms, and pelvis, enabling measurement of postural dynamics and joint angles throughout the surgical procedures without interfering with surgeons performing the surgery. The remaining six participants were fitted with the APDM system, which utilized three IMU sensors placed on the forehead, sternum, and pelvis. Sensor placement for both systems followed the manufacturer's guidelines. Before each surgical session, participants performed a standardized N-pose calibration (Xsens Technologies, 2025) to initialize the IMU systems and align the sensor with reference positions.

To minimize variability, a baseline normalization procedure was conducted before measuring (or surgeries), and all maximum voluntary electrical activation (MVE) tasks in the calibration were administered by the same research team member. Following EMG sensor placement, participants stood in a relaxed posture for 30 s to record baseline muscle activity. The MVE protocol included two individual tasks, each asked to be held for 10 s with recorded start and end times. For the shoulder MVE, participants performed a maximal shrug combined with an upward arm pull against static resistance, targeting the upper trapezius muscles. For the lumbar MVE, participants executed a trunk extension against resistance to elicit maximal activation of the lumbar paraspinals (Cha et al., 2024). This protocol aligns with established EMG normalization practices, which recommend isometric exertions of 8–10 s to ensure reliable peak muscle activation for both amplitude and frequency analyses (McGill, 1991). A 1-min rest between MVE tasks was provided to ensure recovery and prevent potential injuries.

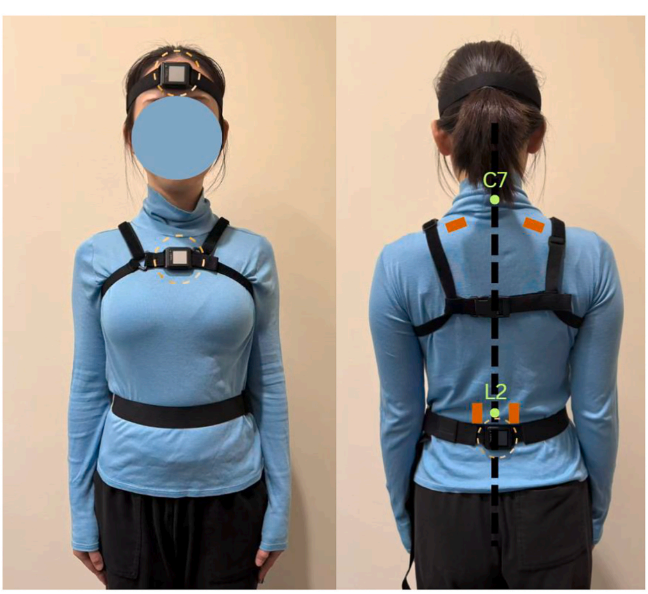

Fig. 2. Locations of EMG and APDM IMU sensors. The C7 and L2 vertebrae served as spinal reference points. Orange rectangles indicate EMG sensor positions, which were placed directly on the skin beneath the clothing. The straps and OPAL units (marked with yellow dashed circles) represent the IMU sensors.

2.5. Data analysis

2.5.1. EMG and IMU data analysis

Biomechanical and physiological data collected from EMG and IMU sensors were analyzed to derive metrics including percent of maximum voluntary electrical activation (%MVE), median power frequency (MPF), and neck joint kinematics. Where applicable, outcome variables were normalized to individual baseline values to control for inter-subject variability. To eliminate potential confounding effects of surgical duration on muscle activation and fatigue measures, only the first hour of data was included in the analysis since it was the shortest procedure length.

EMG signals were visually inspected for quality before analysis. Trials showing continuous high-amplitude interference or flat-line saturation, indicative of electrode detachment or severe noise, were excluded from further processing. A 20–300 Hz band-pass filter was applied to EMG signals to remove motion artifacts and high-frequency noise while preserving the physiologically relevant EMG spectrum (Noraxon USA Inc, 2021; De Luca et al., 2010), together with a notch filter at 60 Hz and its harmonics to eliminate environment interference. Baseline muscle activity was quantified by calculating root mean square (RMS) values during a 30-s resting period, and baseline noise was subsequently subtracted from RMS values obtained during the surgical tasks (Hansson et al., 2000). For MVE computation, the highest RMS amplitude observed during the 10-s isometric contraction task was extracted for each muscle. EMG signals recorded during surgery were then processed using a moving RMS with a 0.5-s shifting window and normalized to their respective MVEs to yield %MVE values (Jonsson, 1982). To characterize muscle activation intensity throughout the surgical procedure, the 10th, 50th (median), and 90th percentiles of %MVE were calculated. In addition, the median power frequency (MPF), a common index of localized muscle fatigue (Wang et al., 2018), was computed using built-in MATLAB functions (R2024a, MathWorks Inc., Natick, USA) with a 0.125-s sliding window and 0.0625-s overlap, following established procedures (McDonald et al., 2019; Muceli and Merletti, 2024). Because live surgeries cannot be interrupted to obtain reference contractions or fixed postures, a fixed 60-min segment was analyzed for each participant to control exposure duration. MPF values were calculated at 10% increments across the surgical task (Keller et al., 2018). To capture fatigue-related spectral shifts, we compared the average MPF between the first and last 10% (Cunningham et al., 2016) of the selected 1-h surgical procedure. Both the absolute difference (ΔMPF, Hz) and the relative change (% of the mean MPF across the full task) were reported.

IMU quaternion data from the head and chest were filtered using an exponential moving average filter ($\alpha = 0.1$). The orientation of the calibration posture was calculated over a 1-min standing period and represented as quaternions. Relative head-to-chest orientation was derived through quaternion operations and subsequently converted into Euler angles (roll, pitch, yaw). Deviations in Euler angles from baseline were computed to quantify postural changes during surgery. Additional kinematic metrics included the percentage of time spent in static posture (defined as angular velocity <1°/s) (Yu et al., 2017) and the range of motion (ROM) for each axis, calculated as the difference between the 95th and 5th percentiles of the relative Euler angles. In addition, postural stability during surgery was examined to evaluate whether changes in body posture could confound the interpretation of EMG-derived fatigue metrics (Appendix A5).

All EMG and IMU data analysis and visualization were performed in MATLAB (R2024a, MathWorks Inc., Natick, USA).

2.5.2. Statistical analysis

Prior to statistical modeling, all continuous outcome variables were screened for distributional assumptions using a combination of Shapiro-Wilk tests and visual inspection of histograms, Q-Q plots, and model residuals. As an initial exploratory analysis of the overall effect of control and exoskeleton conditions, three multivariable analysis of variance (MANOVA) models were applied to EMG-MVE, EMG-MPF, and IMU metrics (Appendix A4). Metric-specific linear mixed-effects models were fitted for individual continuous outcomes to facilitate physiologically interpretable inference. These models included condition (control vs. exoskeleton) as a fixed effect and participant ID as a random intercept to account for within-subject dependency. Dependent variables included changes in subjective discomfort (post-pre surgery), normalized muscle activity (%MVE), decreases in median power frequency (MPF), mean joint angles, and ranges of motion (ROM). For ordinal survey outcomes with three response levels , cumulative link mixed models (CLMMs) with a logit link were applied, with participant ID included as a random effect. As the observed differences only spanned three levels (-1 = decrease, 0 = no change, $+1$ = increase), these were treated as a three-level ordinal response.

To control for multiple testing across questionnaire items and outcome-specific models, the Benjamini-Hochberg procedure was applied to the overall p-values. Statistical significance was defined as an adjusted $p < 0.05$. All statistical analyses were conducted in R (version 4.5.2; Posit PBC, Boston, USA).

3. Results

3.1. Collected data overview

The 36 surgical procedures included both orthopedic and soft tissue procedures, with a mean duration of 2.36 h (SD = 0.80 h). Of these surgical procedures, 18 were conducted under the control condition and 18 under the exoskeleton intervention condition. In all procedures, participants acted as the primary operating surgeon.

3.2. Self-reported discomfort

Participants' self-reported discomfort in four different upper-body regions (i.e., left shoulder, right shoulder, neck, and back) based on their pre-surgical and post-surgical surveys (Table 2). The exoskeleton group has no procedures with increases in neck stiffness and neck pain. However, the exoskeleton group has more procedures with increased left and right shoulder pain. Statistical analysis of symptom score changes before and after surgery revealed that participants in the exoskeleton group experienced significantly smaller increases in left shoulder stiffness ($p = 0.04$), right shoulder stiffness ($p = 0.04$), neck stiffness ($p = 0.001$), and neck pain ($p = 0.04$) compared with the control group.

Table 2
Counts of surgical procedures with decrease, no change, or increase in self-reported discomfort in four upper body regions based on pre-surgical and post-surgical surveys. Significant variables are marked with *.

$n = 36$		Control	Exoskeleton
Left shoulder/arm stiffness*	Decrease	0	1
	No change	14	16
	Increase	4	1
Right shoulder/arm stiffness*	Decrease	0	2
	No change	14	15
	Increase	4	1
Left shoulder/arm pain	Decrease	0	0
	No change	17	16
	Increase	1	2
Right shoulder/arm pain	Decrease	0	0
	No change	17	16
	Increase	1	2
Neck stiffness*	Decrease	0	4
	No change	12	14
	Increase	6	0
Neck pain*	Decrease	0	1
	No change	15	17
	Increase	3	0
Back stiffness	Decrease	0	2
	No change	15	13
	Increase	3	3
Back pain	Decrease	0	2
	No change	15	14
	Increase	3	2

3.3. EMG results

3.3.1. Muscle activation

The EMG results include 25 surgical procedures (15 control, 10 exoskeleton) due to sensor unavailability, signal noise, or technical failures (e.g., battery depletion, dropouts, loose connections). No statistically significant differences in muscle activity (%MVE) were observed between the control and exoskeleton conditions across all assessed muscle groups. However, the exoskeleton group trended lower group-average activation of the left and right lumbar paraspinals, but the trend is not statistically significant. Relatively large variability was observed in %MVE values across both conditions, as reflected by the relatively large standard deviations in relation to the means, particularly in the trapezius muscles (Appendix A1).

3.3.2. Muscle fatigue

In the left upper trapezius, the control group exhibited a larger decrease ($p < 0.01$) in MPF (mean = 3.7 Hz, SD = 4.3) than the exoskeleton group (mean = -0.5 Hz, SD = 1.9). When expressed as a percentage of mean MPF, the reduction in the control group (mean = 5.1%, SD = 4.7) was also significantly greater than in the exoskeleton group (mean = -1.0%, SD = 2.8; $p < 0.002$) (Fig. 3).

Similar patterns were observed in the right upper trapezius, where the MPF decrease was significantly greater ($p < 0.002$) in the control group (mean = 2.7 Hz, SD = 3.8) compared to the exoskeleton group (mean = -0.3 Hz, SD = 2.5). The percent decrease in MPF (mean = 3.5%, SD = 5.0 vs. mean = -0.6%, SD = 3.9) was also statistically significant ($p < 0.05$) (Fig. 3).

In contrast, no significant differences were found in the lumbar paraspinals between conditions. MPF changes in the left and right lumbar paraspinals were small and highly variable across participants.

3.4. Joint motion

3.4.1. Mean neck kinematics

The IMU results include 25 surgical procedures (12 control, 13 exoskeleton) due to sensor unavailability, signal noise, or technical failures (e.g., battery depletion, dropouts, loose connections). No statistically significant differences were observed in mean neck motion metrics between the control and exoskeleton conditions. However, the exoskeleton group (mean = 14.7°) exhibited a trend toward decreased mean neck flexion angles than the control group (mean = 19.7°) ($p = 0.10$).

3.4.2. Percentage of time in static postures

No statistically significant difference was observed in the percentage of time spent in static neck postures between the control and exoskeleton

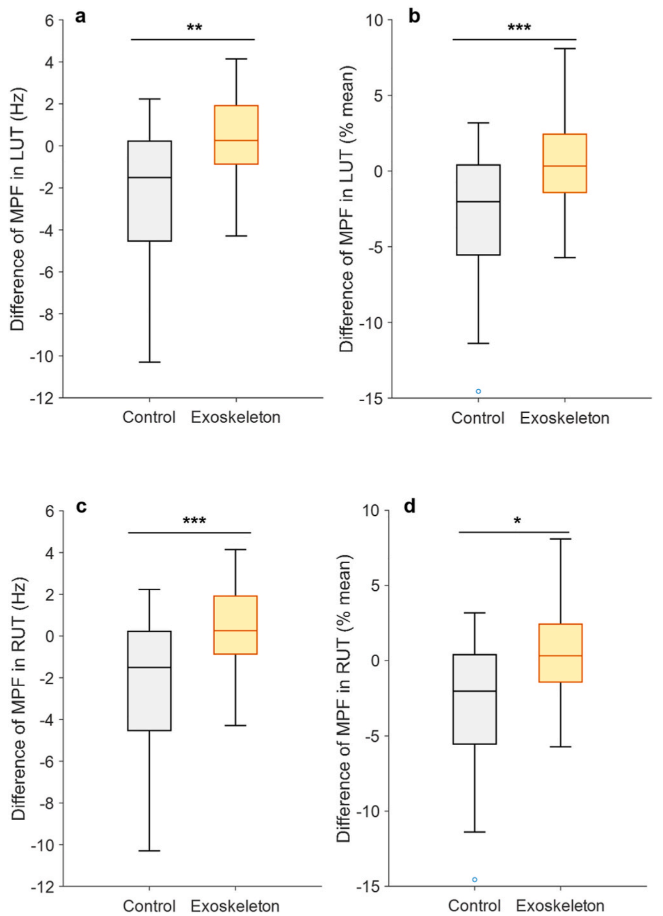

Fig. 3. Box plots of the left upper trapezius (LUT; a and b) and right upper trapezius (RUT; c and d) post-pre median power frequency (MPF) differences. In each box, the horizontal line denotes the median, and the top and bottom edges correspond to the 75th and 25th percentiles. Significant differences are indicated with * if $p < 0.05$, ** if $p < 0.01$, and *** if $p < 0.002$.

groups.

3.4.3. Range of motion

The ROM of neck flexion/extension and neck lateral bending in the exoskeleton group was 10 and 26° lower ($p < 0.05$) than that in the control group (Fig. 4).

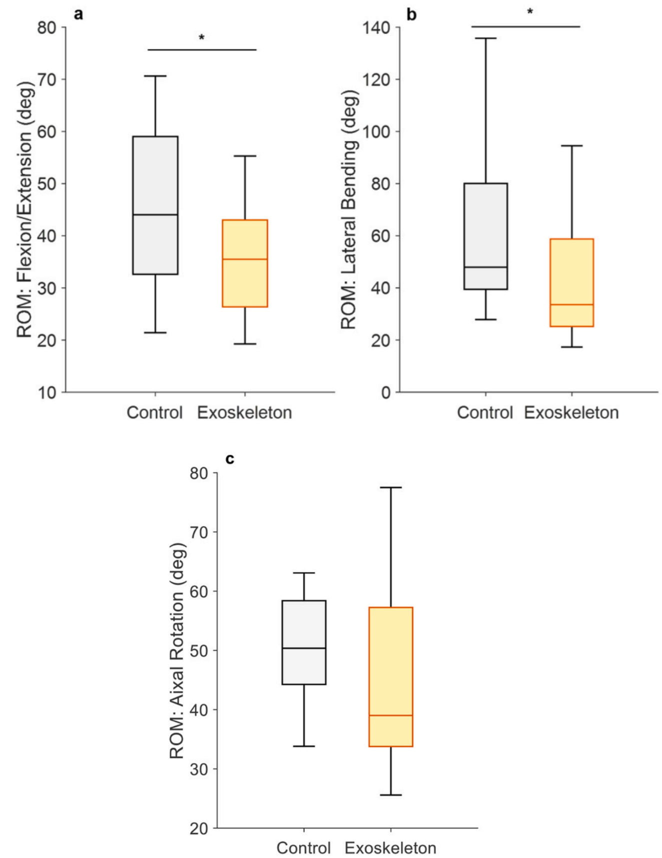

Fig. 4. Box plots of the range of motion in three dimensions: flexion/extension (a), lateral bending (b), and axial rotation (c) in degrees. Significant differences are indicated with * if $p < 0.05$.

4. Discussion

4.1. Questionnaire responses

The questionnaire responses showed that the exoskeleton may help mitigate shoulder and neck discomfort after the surgery, showing significant decreases in reported shoulder stiffness, neck stiffness, and neck pain in comparison to controls. Prior survey research of 3174 veterinarians reported that MSS were most commonly in the neck (66.6%) and shoulders (60.5%), with these regions also producing the greatest levels of physical disability in daily functional tasks (28.7% and 29.5%, respectively) (Kozak et al., 2014). The alignment between the questionnaire findings and the broader evidence highlights the impact of a forward neck posture exoskeleton targeting the neck and shoulder regions. Since MSS in these areas are both highly prevalent and strongly associated with functional impairment, interventions such as posture-supporting exoskeletons may offer meaningful ergonomic benefits.

4.2. Muscle fatigue and muscle activity

A downward shift in the EMG frequency spectrum, particularly reflected in mean and median power frequency, is associated with muscle fatigue (De Luca, 1984; Merletti et al., 1994). This shift primarily arises from a reduction in muscle fiber conduction velocity during sustained activity (Kupa et al., 1995). Decreases in mean or median power frequency have been established as reliable indicators of localized muscle fatigue (Merletti et al., 2001; González-Izal et al., 2012). More recently, these spectral measures have also been applied in surgical ergonomics research to monitor muscle fatigue in operative settings (Merbah et al., 2023). In the present study, we found that MPF decreases were significantly smaller in the exoskeleton group compared to the control group, suggesting that the exoskeleton mitigated the development of muscle fatigue. This effect was most pronounced in the upper trapezius muscles: while the control group exhibited marked declines in MPF in both the left and right upper trapezius, indicative of fatigue, the exoskeleton group showed minimal changes. These findings further support the interpretation of MPF decline as a reliable marker of muscle fatigue and demonstrate the potential of exoskeleton use to reduce muscle fatigue during surgery. Our findings align with prior research reporting that upper-body exoskeletons can mitigate muscular demands during static postures by providing mechanical support (Bär et al., 2021). One possible mechanism underlying the observed reduction in muscle fatigue is the biomechanical redistribution of load facilitated by the passive exoskeleton's headgear and nitinol rod assembly. By transferring a portion of the moment generated by head weight through the external frame, the device lessens the burden on the cervical spine and the upper trapezius muscles, reducing the need for continuous muscle activation and thus delaying fatigue onset. This mechanism finds support in biomechanics literature. De Looze et al. (2016) highlighted that passive exoskeletons can reduce physical loading by providing external support that redistributes internal joint moments to the device structure. The lack of significant effects in the lumbar paraspinals suggests that the exoskeleton's influence on transferring load from neck to back does not necessarily have a negative impact on back muscles, which is consistent with its intended design targeting the upper body. In summary, the study showed muscle evidence of the impact of the passive neck-support exoskeleton.

Interestingly, while the exoskeleton was primarily designed to support the neck-shoulder regions, it did not lead to significant reductions in trapezius muscle activation. Several factors may account for this outcome. One possibility is that novice users, lacking prior experience with the device, continued to activate neck and shoulder muscles out of habit or in response to the support provided by the exoskeleton, thereby counteracting its intended mechanical unloading effect. This may have occurred despite participants having up to two procedures working with the intervention. This interpretation aligns with findings from Flor-Unda et al. (2023) and Vallée (2024), which suggest that an adaptation period is often required when using exoskeletons, during which users may initially show altered muscle recruitment patterns and even increased discomfort. Variability in sensor placement, %MVE calibration (which was calibrated for each case), and individual posture strategies may also have contributed to the relatively high standard deviations observed in trapezius %MVE values, further obscuring any potential effect.

Although no statistically significant differences were observed in mean muscle activity between conditions, several factors may explain this outcome. First, muscle activation during surgery exhibited large inter- and intra-individual variability, reflecting differences in surgical duration, technique, and surgeon posture across and within procedures. Such variability can mask modest but consistent reductions in muscular load. Second, the exoskeleton's benefit may have occurred primarily at the extremes of the activation spectrum, by reducing short periods of high muscle load (above the 90th percentile) and increasing resting time (below approximately 0.5% MVE), which were not captured in the analyzed percentile ranges. Finally, the largest unloading effect likely occurred in the neck-shoulder region, which bears the highest static demand during surgery. This interpretation aligns with the significant reduction in fatigue observed in the trapezius muscles, indicating that the exoskeleton effectively mitigated localized neck loading even when overall mean activity levels remained statistically similar.

4.3. Joint motions

In this study, the exoskeleton group exhibited a trend toward reduced mean neck flexion angles compared to the control group. Both values are close to the flexion angles of 17° in a research by Asadi et al. (2022) on veterinary surgeries. However, the control group has a mean neck flexion angle of 19.7°, which is considered ergonomically risky. According to the Rapid Upper Limb Assessment (RULA), neck flexion of 10–20° corresponds to a score of 2, indicating a moderate risk level (McAtamney and Corlett, 1993). Thus, while neither condition achieved a fully neutral posture, the exoskeleton group-maintained surgeons closer to the lower end of this category (having a mean neck flexion angle of 14.7°), suggesting a modest ergonomic advantage.

The exoskeleton significantly reduced the range of motion (ROM) in neck flexion and lateral bending, while preserving the mean head posture. This indicates that the device did not alter typical working head positions but instead acted as a stabilizer, limiting deviations from the preferred posture more than in the control condition. Such stabilization can be ergonomically beneficial, as reducing excursions into large postural angles is known to decrease mechanical strain on the cervical spine and shoulder muscles, thereby lowering the risk of discomfort, fatigue, and musculoskeletal disorders during prolonged static tasks (Punnett and Wegman, 2004; Bernard and Putz-Anderson, 1997).

At the same time, restricting ROM may also introduce disadvantages. Constraining natural movement could limit postural variation, both of which are important for reducing static load and preventing localized fatigue (Visser and van Dieën, 2006). Thus, while the exoskeleton appears to mitigate potentially harmful postural extremes, future work should consider the balance between stabilization and maintaining sufficient mobility to support healthy movement variability. In surgical practice, the goal is a device that tempers excessive flexion or lateral bending while preserving sufficient ROM across the relevant axes and planes, so surgeons can still adopt necessary postures when required.

Other researchers have also explored the relationship between posture control and fatigue reduction. For instance, Kim et al. (2018) found that exoskeletons could reduce neck and shoulder loading in tasks involving overhead work, partially through posture stabilization. While their context involved more dynamic tasks, our findings extend these insights to static, low-variability settings like surgery, suggesting that postural stabilization alone can yield fatigue-related benefits.

4.4. Veterinary surgeries and human surgeries

Previous ergonomics studies of human surgeons have consistently documented high static and ergonomically risky postures of the neck and trunk during surgical tasks, contributing to elevated musculoskeletal strain. Using IMUs, Norasi et al. (2021, 2023) quantified intraoperative neck and trunk kinematics in vascular surgeons and reported sustained neck flexion exceeding 20° for more than half of the operative time, along with asymmetric trunk postures and limited opportunities for recovery. Also, previous muscle activity studies in surgeons have found elevated cervical erector spinae and upper trapezius activity during open surgery compared with minimally invasive approaches (Szeto et al., 2010), and fatigue-related declines in lumbar erector spinae during thoracoscopic procedures (Yoon et al., 2016). Similarly, in this study, when no exoskeleton was used, we found a near 20-degree neck flexion, and EMG evidence also shows lumbar muscles have signs of fatigue, which is consistent with previous surgical ergonomic studies. Interestingly, unlike human surgeries, our results, together with those of Asadi et al. (2022), who reported mean neck flexion angles of 19.7° and 17° in veterinary procedures, show smaller neck motion magnitudes than those typically observed in human surgical studies (approximately 30–40°). This finding contrasts with the common assumption that veterinary surgeries involve more awkward and variable postures due to differences in animal size and positioning. However, this apparent contradiction may be explained by the dynamic nature of veterinary surgical workflows. Veterinary surgeons often move around the operating table, adjust their stance, and reposition the patient or instruments throughout the procedure. In contrast, human surgeons, particularly in vascular or minimally invasive operations, typically maintain fixed visual and hand positions relative to a static field, leading to prolonged, low-amplitude cervical flexion and higher cumulative musculoskeletal loading (Norasi et al., 2021, 2023). Thus, while veterinary surgeries are more variable in posture, their intermittent and mobile working patterns may impose less continuous static strain on the neck and trunk compared with the highly constrained environments of human operating rooms.

Although this study examined veterinary surgical tasks, its implications extend to understanding how passive exoskeletons may function in human surgery. In our study, the neck-support exoskeleton reduced trapezius muscle fatigue during prolonged procedures, indicating its potential to mitigate localized fatigue under static postural load. Comparable benefits have been reported in human surgical contexts, where passive back- and arm-support exoskeletons decreased trunk flexion, reduced perceived workload, and improved postural endurance during open and vascular operations (Norasi et al., 2021, 2023; Wang et al., 2025; Cha et al., 2024). However, several biomechanical distinctions influence how exoskeletons function across these settings. Veterinary surgeons often perform open procedures on animals of varying sizes and table heights (Villarreal et al., 2023), requiring various body postures to maintain visual access to the operative field (Seagren et al., 2022). In contrast, human surgeons typically work in standardized operating rooms, where posture is more constrained but prolonged, particularly in vascular or minimally invasive procedures that demand sustained neck and shoulder elevation with fine motor precision. Consequently, veterinary applications primarily challenge exoskeletons' flexibility to accommodate dynamic, asymmetric postures, whereas human surgical contexts emphasize static load redistribution for endurance support. Nonetheless, systematic comparative studies between veterinary and human surgeries are needed to quantify differences in postural variability, muscle loading, and exoskeleton effectiveness across surgical domains.

4.5. Technical considerations

Previous studies have highlighted the ergonomic risks in surgery, emphasizing force, repetition, and posture as key contributors to work-related musculoskeletal disorders (MSDs) (Kilbom, 1994; Occhipinti, 1998). Surface electromyography (EMG) and inertial measurement units (IMUs) have increasingly been used in ergonomic research to provide objective, real-time assessments of muscle activation, fatigue, and posture. Prior work has demonstrated the convenience and feasibility of using these tools in surgical settings. For example, Asadi et al. (2018) used IMUs to examine upper extremity postures in veterinary surgeons, Fan et al. (2022, 2024) combined both muscle activity and joint motion data to comprehensively assess surgeons' workloads and movements. Two IMU sensors were used across different sessions due to equipment availability, but both shared identical sensing mechanisms, recording three-axis acceleration and angular velocity at comparable sampling rates and resolutions. Each sensor was calibrated and verified through identical static and dynamic trials before data collection. Although no prior studies have directly compared these IMU models, our analyses focused on relative within-trial metrics (e.g., mean neck angle, temporal changes) rather than absolute values, minimizing potential cross-sensor bias.

4.6. Limitations and future works

The study has several limitations. First, of the 36 surgical cases enrolled (all with surveys), 25 (69%) yielded analyzable EMG/IMU data, while 11 (31%) were excluded due to sensor unavailability, signal noise, or technical failures (e.g., battery depletion, dropouts, loose connections). Analyses were thus conducted on a complete-case basis. Although these losses were technical rather than participant-related, longer or more demanding procedures may have been more prone to sensor failure, potentially biasing results toward more stable cases. Second, the placement of surface EMG and IMU sensors may have experienced slight shifts over the course of the procedures due to the prolonged duration of surgical tasks and the frequent movements required, potentially affecting the accuracy of signal acquisition. To minimize potential sensor displacement, all sensors were securely attached to each participant's body while ensuring no noticeable discomfort. Third, the type and complexity of surgical procedures were not experimentally controlled, introducing variability in physical demands that could influence muscle activation, posture, and fatigue independently of the intervention. Unlike simulated tasks, real surgeries involve natural microbreaks-brief moments of rest, such as waiting for an instrument or scope change. These unscheduled pauses can temporarily reverse or obscure fatigue-related EMG trends (Öberg et al., 1991), resulting in variable or even increased MPF values (Madeleine et al., 2003). Additionally, the low-load, static nature of the tasks may lead to inconsistent fatigue development and reduce the reliability of MPF as a fatigue indicator in this setting. Also, MPF is known to be influenced by factors beyond fatigue, such as posture, muscle length, and contraction intensity. In the present study, several methodological controls were implemented to mitigate these sources of variability. The analysis was confined to a standardized 60-min window across all participants to ensure comparable exposure duration. MPF values were further normalized to each participant's overall mean to reduce interindividual differences and task-related variations. Together, these steps were intended to enhance the robustness of MPF as an indicator of time-dependent muscle fatigue during live surgical procedures.

Another methodological limitation of this study is that surface EMG was recorded from the upper trapezius and lumbar paraspinals, but not from the cervical erector spinae (CES) muscles, such as the splenius capitis. While CES would provide the most direct assessment of neck extensor activity, several practical constraints prevented its inclusion in this field study. There is currently no consensus on optimal electrode placement for CES, which consists of small, deep muscles interlaced with adjacent neck musculature, resulting in a high likelihood of signal crosstalk. In addition, the cervical region presents large inter-individual variability in hair density, soft-tissue thickness, and limited surface area for electrode adhesion, which often compromises data quality and can interfere with medical devices or sterile coverings during surgery. In contrast, the upper trapezius was selected because it is widely used as a reference muscle in neck and shoulder ergonomics research and has established action levels for musculoskeletal risk assessment (Arvidsson et al., 2021). Also, the upper trapezius functions primarily as a stabilizing and co-activated muscle during sustained neck-shoulder postural tasks like surgeries, showing comparable activation across cervical flexion and extension despite not acting as a prime mover (Lecompte et al., 2007). Moreover, trapezius activity and fatigue are well-established indicators of neck-shoulder musculoskeletal disorder risk, making it a meaningful target for ergonomic evaluation of assistive devices in long-duration static tasks (Westgaard et al., 2001; Balogh et al., 2019). The inclusion of the lumbar paraspinals served primarily as a comparative or "negative control" muscle, allowing us to examine whether the exoskeleton's back frame introduced unintended load redistribution to the lower back. Although minimal changes were expected in LES activity, its inclusion provided a reference to ensure that reductions in neck loading were not offset by increased lumbar effort. Future studies could incorporate cervical erector spinae measurements using refined placement protocols or intramuscular EMG to provide a more direct quantification of neck extensor loading under exoskeleton assistance.

Future studies should aim to increase sample size, standardize surgical conditions, and employ real-time monitoring to verify sensor stability during extended use. A key step toward improving exoskeleton effectiveness in surgery is accumulating a larger number of clinical cases so that surgeons can become more familiar with the device and its use. Greater exposure may not only reduce initial discomfort or awkwardness but also address concerns about altered muscle activation patterns through adaptation over time. At the same time, the barriers to exoskeleton adoption are complex, involving human factors, workflow integration, and user acceptance (Cha et al., 2020). For a neck-support exoskeleton to provide sustained effectiveness in surgical practice, critical aspects such as long-term comfort, ease of donning and doffing, compatibility with surgical tasks, and potential impacts on surgical ergonomics must be carefully considered.

In summary, the results suggest that a passive exoskeleton may help reduce neck-shoulder muscle fatigue during real-world veterinary surgeries. However, its potential to restrict movement in certain directions highlights the need for careful consideration and practical evaluation before clinical implementation.

CRediT authorship contribution statement

Bowen Zheng: Writing – original draft, Visualization, Methodology, Formal analysis, Data curation. **Xuelong Fan:** Writing – review & editing, Validation, Methodology, Formal analysis. **Sun Young Kim:** Writing – review & editing, Supervision, Resources. **Denny Yu:** Writing – review & editing, Supervision, Resources, Project administration.

Declaration of generative AI and AI-assisted technologies in the writing process

During the preparation of this work, the authors used ChatGPT in order to improve grammar and language quality. After using this tool/service, the authors reviewed and edited the content as needed and take full responsibility for the content of the publication.

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Declaration of competing interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

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Appendix

Table A.1
Summary of MVE and MPF results of 25 surgical procedures. Hedges' *g* values represent standardized mean differences computed as the experimental condition minus the control condition; negative values indicate lower values under the experimental condition relative to the control condition, whereas positive values indicate higher values under the experimental condition.

	Control		Exoskeleton		*p*-value	Hedge's *g*
	Mean	(SD)	Mean	(SD)		
%MVE						
Left Upper Trapezius						
10th	4.3	(2.9)	4.9	(4.3)	0.46	0.30
50th	12.3	(10.7)	12.9	(10.0)	0.99	−0.10
90th	31.6	(13.3)	26.9	(18.1)	0.79	−0.22
Right Upper Trapezius						
10th	4.2	(3.1)	5.3	(4.8)	0.76	−0.06
50th	14.1	(17.0)	12.5	(8.1)	0.90	−0.17
90th	34.7	(30.2)	24.9	(13.1)	0.50	−0.29
Left Lumbar Paraspinals						
10th	5.4	(6.1)	4.9	(3.5)	0.66	−0.08
50th	13.0	(12.9)	11.4	(5.5)	0.49	−0.15
90th	24.8	(18.8)	19.3	(5.2)	0.15	−0.39
Right Lumbar Paraspinals						
10th	5.7	(5.2)	4.6	(3.1)	0.31	−0.33
50th	13.4	(3.9)	12.5	(8.6)	0.41	−0.29
90th	25.2	(17.8)	20.1	(6.2)	0.14	−0.39
MPF						
Left Upper Trapezius						
MPF (first 10%, Hz)	77.1	(12.4)	70.9	(12.6)	–	–
MPF (last 10%, Hz)	73.4	(10.3)	71.4	(10.2)	–	–
MPF decrease (Hz)	3.7	(4.3)	−0.5	(1.9)	<0.01*	−1.12
MPF decrease (% mean)	5.1	(4.7)	−1.0	(2.8)	<0.002*	−1.43
Right Upper Trapezius						
MPF (first 10%, Hz)	76.4	(13.1)	69.8	(12.9)	–	–
MPF (last 10%, Hz)	73.7	(11.5)	70.1	(11.2)	–	–
MPF decrease (Hz)	2.7	(3.8)	−0.3	(2.5)	<0.002*	−0.85
MPF decrease (% mean)	3.5	(5.0)	−0.6	(3.9)	<0.05*	−0.84
Left Lumbar Paraspinals						
MPF (first 10%, Hz)	81.6	(5.9)	84.6	(2.5)	–	–
MPF (last 10%, Hz)	79.7	(3.6)	82.3	(4.2)	–	–
MPF decrease (Hz)	1.9	(3.7)	2.3	(4.9)	0.81	−0.08
MPF decrease (% mean)	2.4	(4.3)	2.2	(5.3)	0.97	−0.04
Right Lumbar Paraspinals						
MPF (first 10%, Hz)	89.9	(7.7)	84.7	(2.4)	–	–
MPF (last 10%, Hz)	86.6	(8.9)	83.1	(9.8)	–	–
MPF decrease (Hz)	3.3	(4.9)	1.6	(6.0)	0.41	−0.31
MPF decrease (% mean)	3.5	(5.3)	1.6	(5.9)	0.36	−0.34

Table A.2
Questionnaire

Pre-Surgical Discomfort			
Area/subject	No	Slight	Substantial
Shoulder/arm stiffness	L |R	L |R	L |R
Shoulder/arm pain	L |R	L |R	L |R
Neck stiffness			
Neck pain			
Back stiffness			
Back pain			
Post-Surgical Discomfort			
Area/subject	No	Slight	Substantial
Shoulder/arm stiffness	L |R	L |R	L |R
Shoulder/arm pain	L |R	L |R	L |R
Neck stiffness			
Neck pain			
Back stiffness			
Back pain			

Table A.3
Statistical results for the 10th/50th/90th percentile of neck joint motions between control and exoskeleton group

Motion type		*p*-value
Flexion/Extension	10th	0.32
	50th	0.12
	90th	0.10
Lateral bending	10th	0.43
	50th	0.15
	90th	0.37
Axial rotation	10th	0.44
	50th	0.93
	90th	0.38

Table A.4
Overall control/exoskeleton group differences of MVE, MPF, and joint motion metrics from MANOVA. For each outcome domain (MPF, MVE, and IMU joint kinematics), a one-way multivariate analysis of variance (MANOVA) was first conducted to examine the overall effect of experimental condition (control vs. exoskeleton) on the set of correlated dependent variables. Hedge's *g* was computed for each individual dependent variable, comparing the control and exoskeleton conditions (exoskeleton minus control). To summarize the overall magnitude of condition effects within each outcome domain, the absolute Hedge's *g* values were averaged across variables.

Metrics group	*p*-value	Hedge's *g*
%MVE	0.88	0.23
MPF	0.03	0.62
Joint motions	0.03	0.33

Table A.5
Posture analysis of the first 10% and last 10% of the 60-min time of interest. As an additional analysis, posture changes between the first and last 10% of the surgical procedure were evaluated to examine potential posture-related confounds in EMG-derived fatigue measures. Posture outcomes were analyzed using linear mixed-effects models, with time (first vs. last 10%) and experimental condition (control vs. exoskeleton) as fixed effects, and participant ID included as a random intercept. Analyses were conducted separately for flexion/extension, axial

rotation, and lateral bending. Model-estimated marginal means were used to quantify pre-post posture changes within each condition.

Posture dimension	*p*-value	Estimate
Flexion/extension	0.89	−0.13
Lateral bending	0.53	1.12
Axial rotation	0.31	−2.3
Overall	0.71	−0.44

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