A Bioengineering Investigation of Cervical Collar Design and Fit: Implications on Skin Health

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**Keywords**

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Abstract

*Background:* Cervical collars restrict cervical spine movement to minimise the risk of spinal cord injury. Collars apply mechanical loading to the skin putting it at risk of skin damage. Indeed, cervical collar-related pressure ulcers are unacceptably prevalent, especially at the occiput, mandibles, and chin. Collar design and fit are often key considerations for prevention.

*Methods:* This comprehensive study evaluated four commercial prehospital and acute care cervical collars. Pressure, microclimate, transepidermal water loss and skin hydration were measured at the interface between the device and the skin. Range of motion restriction was measured to evaluate effective immobilisation. Head, neck, and shoulder morphology was evaluated using three-dimensional scans.

*Findings:* The occiput experienced significantly higher interface pressures than the chin and mandibles for most collar designs. Interface pressure at the occiput was significantly higher for the Stiffneck extrication collar compared to the other collar designs. The Stiffneck collar also provided the most movement restriction, though not significantly more than other designs. Relative humidity at the device skin interface was significantly higher for the Stiffneck and Philadelphia collars corresponding to closed cell foam padding, in contrast to the open cell foams lined with permeable fabric used in the other collars. Collar discomfort correlated with both occipital pressure and skin humidity.

*Interpretation:* The occiput is at increased risk of cervical collar-related pressure ulcers during supine immobilisation, especially for Stiffneck extrication collars. Lined open-cell foams could be used to minimise skin humidity and increase comfort.

# Introduction

Cervical collars are routinely used to immobilise trauma patients with a suspected spinal injury (Ahn et al., 2011). Immobilisation reduces the risk of further damage to the spine that could lead to motor and sensory impairment and morbidity in the most severe cases. Collars are worn until spinal clearance can be obtained through clinical examination, often including radiological assessment (National Institute for Health and Care Excellence [NICE], 2016). Cervical collars are also used during rehabilitation for musculoskeletal disorders of the cervical spine. This includes weakness resulting from neuromuscular or musculoskeletal disease, as well as from surgical interventions. Immobilisation is achieved through rigid support around the cervical spine with adjustable straps and height adjustments. The rigid supports and strapping often lead to localised areas of pressure and shear at the device-skin interface (Sparke et al., 2013). The materials that interface with the skin can also contribute to a local increase in microclimate temperature and humidity, adversely affecting the skin’s tolerance to mechanical loads (Kottner et al., 2018). The mechanical loading from interface pressure and shear over time has an associated risk of skin breakdown leading to pressure ulcers (PUs), also termed pressure injury. When caused by medical devices, PUs are also termed medical device-related pressure ulcers (MDRPUs), where skin damage typically conforms with the shape of the overlying device (Black et al., 2010).

The reported incidence of cervical collar-related pressure ulcers varies between observational studies, ranging from 0.7-44% (Ackland et al., 2007; Chan et al., 2013; Davis et al., 1995; Ham et al., 2014; Powers et al., 2006; Wang et al., 2020). Brannigan et al. (2022) report a pooled average of 7%. This high variability is likely a result of differences in preventative measures, setting, and reporting procedures. Several studies suggest under-reporting as a potential limitation in less accessible locations, such as the occiput, where the device and hair may block visible signs of skin damage. Indeed, common sites of skin breakdown include the occiput, chin, shoulders, and clavicle (Ackland et al., 2007). Amongst the risk factors, collar application duration leads to the most significant increase in PU risk (Brannigan et al., 2022). These observational studies do not report the effects of collar design and fit and often do not report the type of collar used.

Experimental studies suggest that different collar designs introduce varied interface pressure, indicating variable PU risks. Studies on interface pressure and range of motion (Tescher et al., 2007; Tescher et al., 2016; Whitcroft et al., 2011; Worsley et al., 2018) demonstrate these differences between collar designs. However, interface pressure alone provides a limited view of ulcer risk and may not directly relate to its ability to restrict cervical movement (Oomens et al., 2010). Several intrinsic and extrinsic factors could contribute to ulceration (Bader & Worsley, 2018). In particular, individual head shape and size may affect collar fit, yet have not been fully evaluated in collar designs. Thus, a more holistic approach, combining biomechanical and physiological assessments, is required to evaluate PU risks from collar usage. A recent study has contributed to this approach, investigating the use of biomarkers to evaluate the skin’s response to loading from cervical collars (Worsley et al., 2018). However, this study was limited to two-collar designs, restricting a full comparison across common designs and materials of collars.

The design and fit of cervical collars in the context of skin health have yet to be widely reported. This study investigated collar function, collar-skin interface pressure, physiological skin response, and patient comfort across four common cervical collar designs in healthy volunteers. A combination of measures previously used to evaluate medical device-related pressure ulcers and anthropometrics was used to compare the effects of different collar designs. The association between collar function, interface pressure, skin response and comfort were also studied.

# Methods

A convenience sample of healthy participants was recruited via poster advertisement from the local university population. Exclusion criteria included age below 18 or above 60 years, illness, infection, skin conditions affecting the neck area, conditions affecting the cervical spine, cardiovascular insufficiencies affecting soft tissue health and recovery, and inflammatory conditions. Informed consent was received from all participants before participation in the study. The study was approved by the University of Southampton Ethics Committee (ERGOII – 18511.A1).

## Test equipment

Four different cervical collars were used for this study (Fig. 1). The Miami J standard collar with sternal pad (Össur UK Ltd, Stockport, UK), made of 2 mm thick LDPE (Low Density Polyethylene) panels with 10 mm thick fabric-lined open-cell foam (Sorbatex). The Stiffneck Select collar (Laerdal Medical Ltd, Orpington, UK), made of 2 mm thick HDPE (High Density Polyethylene) shell with 6 mm thick closed-cell foam padding. The Philadelphia Adjustable Tracheotomy collar (Össur UK Ltd, Stockport, UK), made of 3 mm thick PP (Polypropylene) supports with 10 mm thick formed closed-cell Plastazote foam. The Aspen Vista collar (Aspen Medical Products, Irvine, CA, USA), made of 1-2 mm thick polymer panels with 9 mm thick cotton-lined open-cell foam padding.

Interface pressures were monitored using a commercial system (Tactilus Free Form Sensor System, Sensor Products Inc., Madison, NJ, USA). Their V Series 10x10 mm square sensors have an operating range of 0 and 258 mmHg, reported accuracy of ± 10%, repeatability of ± 2%, and a sampling frequency of 49 Hz (Sensor Products Inc., 2023). Microclimate at the device-skin interface was measured using combined sensors (SHT75; Sensirion AG, Stäfa, Switzerland), recording relative humidity and temperature at 0.5 Hz with report accuracy of ± 1.8% RH and ± 0.3°C, respectively. The skin’s physiological status was assessed by measuring transepidermal water loss (TEWL) and skin hydration. TEWL and skin hydration were measured with commercial sensors Tewameter, MPA9 and Corneometer, MPA9, respectively (Courage & Khazaka, Köln, Germany). Surface scans of individuals’ heads, necks and shoulders were taken using a commercial hybrid structured light scanner in infrared mode with an accuracy of 0.6 mm (EinScan H; Shining 3D, Hangzhou, China). Scans were processed using proprietary software (EXScan H v1.0.5.3; Shining 3D, Hangzhou, China). Range of motion was measured using a handheld digital inclinometer (Digital Levelmeter 1700, SOAR, Japan). With the collars in situ, the participants were also asked to report their perceived discomfort using an 11-point numeric rating scale. The lowest score (0) represents no discomfort at any point, and the highest (10) means extreme discomfort.

## Test protocol

A randomised crossover design was used for this study. Participants were randomised to two of the four commercial collar designs being tested.

Testing was performed in a biomechanics laboratory with a controlled ambient temperature of 20-22°C and relative humidity of 30-40%. Participants were given at least 10 minutes to acclimatise to the laboratory environment before baseline measurements were taken, followed by collar application, intervention data collection, collar removal and post-intervention data collection, as shown in Fig. 2. Participants attended one session, during which two collars were tested according to block randomisation. Participants were asked to have clean, washed skin and have shaved where appropriate at least 48 hours before participation. Demographic data, including age, gender, height, and weight, were collected.

Thereafter, the participant lay supine on a standard viscoelastic mattress for each collar test (Medstrom Ltd, Castle Donington, UK). Baseline physiological measurements (hydration and TEWL) were taken on the underside of the chin (Fig. 3). The collar was then applied according to the manufacturer’s instructions, with feedback on tension provided by the participant. After 10 minutes of application, the interface pressure sensors were inserted between the collar and the skin at the occiput, chin, right, and left mandible (Fig. 3) and a reading of static loading was taken. The pressure sensors continued recording while the range of motion was measured. The participant was asked to perform flexion, extension, and right and left rotation, stopping when they experienced resistance from the collar. Neutral and maximum flexion and extension angles were recorded with the inclinometer resting on the forehead and touching the nose without compressing it (sagittal plane). Maximum right and left rotation were recorded with the inclinometer at 90° resting on the forehead (transverse plane). The pressure sensors were then removed and replaced with the temperature and humidity sensors at the right and left mandible and chin (Fig. 3)—these sampled data for ~30 seconds until a stable recording was achieved. Before collar removal, participants were asked to rank their perceived discomfort using the 11-point verbal rating scale. The collar was then removed, and the physiological skin response measures were taken per the baseline measurements in the same location under the chin (hydration and TEWL). 3D surface scans were then taken of the participants with and without the collar whilst sitting upright. A 10-minute refractory period was imposed between the first collar being removed and baseline measurements being taken for the second collar test to ensure adequate soft tissue recovery.

### Data analysis

Statistical analysis was performed using Python (SciPy library). Data from each test were examined for normality using Shapiro-Wilk and Anderson-Darling tests. Parametric statistics (mean ± standard deviation) were found to be appropriate for analysing hydration and TEWL. Statistical differences between parametric variables were evaluated using a student’s t-test and paired t-test for between-collar and within-collar differences, respectively. Nonparametric statistics (median, interquartile range) were applied to comfort scores, interface pressure, and range of motion. Statistical differences were evaluated using a Wilcoxon signed-rank test for within-collar differences and a Mann-Whitney U test for between-collar differences. Spearman correlation was used to analyse the relationship between non-parametric variables. The statistical significance level was set to p ≤ 0.05 for all outcomes. To account for multiple comparisons performed during the study, a Bonferroni correction was applied to the Alpha value, such that the statistical significance was p < 0.0083 where six pairs of collars were compared.

# Results

## Participant Demographics

A sample of 25 healthy volunteers (10/15, Male/Female) were recruited. The participants had a median age of 21 years (range 18-38), mean height of 173 ± 10 cm, and mean weight of 70.5 ± 16.6 kg, with a corresponding body mass index (BMI) of 23.3 ± 3.3 kg/m2. Table 1 shows the breakdown of these demographics for each collar after randomisation. There were no significant differences between height, weight, and corresponding BMI between collars. However, some distinct differences in gender distribution were noted, with more females testing the Philadelphia and Aspen collars.

## Interface Pressure

Fig. 4 illustrates the interface pressure values across the four measurement sites for each collar during the static lying posture. Significant differences were observed between measurement locations for all collars apart from the Miami J collar, with median pressure values ranging between 51.0 and 66.0 mmHg across the four locations. The Stiffneck collar revealed significantly higher interface pressure at the occiput than the mandibles (p < 0.001), with some readings exceeding the measurable range of the sensors at 258 mmHg. The Philadelphia and Aspen collars also had significantly higher interface pressure at the occiput when compared to the chin (p < 0.008)

There were also significant differences between the collar designs for pressure at the occiput. The occiput pressure of the Stiffneck was significantly higher than the Miami J and Aspen collars (p < 0.008), with a more than two-fold increase compared to the Miami J collar. There were no significant differences between pressures at the chin. However, there were some instances of very high chin pressure for the Miami J and Stiffneck collars and very low or absent chin pressure for the Miami J, Stiffneck and Philadelphia collars. This is evidenced by outliers in the box and whisker plots.

## Cervical Range of Motion

No significant differences in range of motion were observed between the different collars. However, differences in mean value indicate that the Stiffneck collar was the most restrictive, and the Aspen and Miami J collars were the least restrictive. Across all collar designs, the most restriction was observed in the sagittal plane, namely flexion and extension. By contrast, transverse rotation was higher, with mean values just over 2-fold higher than flexion/extension. A high degree of inter-subject variability was observed in all four collars; for example, in left rotation, values ranged from 6.5-54.0° for the Aspen collar.

## Microclimate

Skin temperature and relative humidity did not vary significantly between the three test sites. Significant differences in relative humidity were observed between collars. Mean relative humidities were lowest for the Aspen Vista collar. Relative humidity was significantly increased for the Stiffneck compared to the Miami J (20 %RH mean difference, 95% CI [16,24], p < 0.001 (chin)) and Aspen collars (27 %RH mean difference, 95% CI [23,33], p < 0.001 (chin)). Relative humidity was significantly increased for the Philadelphia compared to the Miami J (15 %RH mean difference, 95% CI [11,20], p < 0.001 (chin)) and Aspen collars (23 %RH mean difference, 95% CI [17,29], p < 0.001 (chin)). Additionally, the Miami J saw increased relative humidity at the chin compared to the Aspen collar (p < 0.008).

Mean skin temperatures were lowest in the Aspen Vista (34.2-34.8 °C) collar and highest for the Philadelphia (35.0-35.5 °C) collar across each measurement site. Though temperatures in the Philadelphia collar were significantly higher than the Aspen Vista collar across at the chin (p < 0.008), the mean differences were small, 0.8°C (95% CI [0.1,1.5]) at the chin, 0.8°C (95% CI [0.3,1.3]) at the right mandible, and 0.6°C (95% CI [0.1,1.2]) at the left mandible. The Stiffneck also experienced significantly higher skin temperatures than the Aspen Vista at the right mandible (p < 0.008), with a mean difference of 0.6°C (95% CI [0.2,1.0]). Across all collars, the mean skin temperatures were above normal physiological ranges for a resting adult male (30-34°C) (Gagge et al., 1967; Winslow et al., 1937).

## Skin Response

There were limited changes in the biophysical skin parameters at the chin, with a high degree of inter-subject variability (Table 3). TEWL and hydration did not differ significantly from baseline (p > 0.5). No significant variations in these measures existed between the different collar designs (p > 0.4). Most TEWL values were within normative ranges (8-14 g/hm2), irrespective of collar type.

## Perceived Discomfort

Fig. 6 shows the perceived discomfort was significantly higher for the Stiffneck than the Miami J and Aspen collars (p < 0.001). Although non-significant, the Stiffneck collar was also more uncomfortable than the Philadelphia collar (p = 0.030). The Stiffneck had a median discomfort score of 6 (range 5-8), compared to Aspen and Miami J, which had median scores of 3 (range 1-5 for both collars).

## Correlation analysis

The relationships between anthropometric measurements and interface pressure were limited (Fig. 7). However, for specific collars, statistically significant correlations between anthropometrics and range of motion were observed. This was particularly the case with the Miami J collar with significant correlations between neck circumference and range of motion in extension-flexion range (r(10) = 0.81, p < 0.01) and transverse rotation range (r(10) = 0.73, p < 0.01).

There was a moderate positive correlation between the perceived discomfort and pressure at the occiput, r(46) = 0.41, p < 0.01. No correlation was observed between the perceived discomfort and other interface pressure measurements. Perceived discomfort also had a moderate positive correlation with the relative humidity at all three measurement sites, r(46) = 0.56 (chin), 0.56 (right), 0.49 (left), p < 0.01. Range of motion measures were most notably moderately negatively correlated with the pressure at the occiput r(46) = -0.48, -0.50 (extension, rotation), p ≤ 0.001.

# Discussion

This study compared the biomechanical, physiological, and perceptual responses between four cervical collar designs in healthy volunteers. The interface conditions varied significantly between collars and between locations on the collar, implying effects from collar designs and materials used within each device. The effectiveness of restricting movement in the cervical spine varied between collars and depended on the plane of motion. Pressure and relative humidity values observed in this study for specific collar conditions could put skin and underlying soft tissues at risk of damage if sustained for prolonged periods. The results from this study could inform the design of new collars that combine effective movement restriction with optimised skin health for vulnerable patients.

Interface pressures were significantly higher at the occiput than at the mandible and chin for the Stiffneck, Philadelphia and Aspen collars. This agrees with the literature indicating an increased incidence of pressure ulcers at the occiput (Powers et al., 2006). The Stiffneck had the highest median occipital interface pressure (155, IQR = 122,258 mmHg), which could be associated with its limited conformity to the shape of the head, especially around the rear portion of the collar. This increased interface pressure for the Stiffneck collar aligns with the reported high incidence of pressure ulcers using this design of extrication collar (Ham et al., 2016). By contrast, the Miami J had very similar and lower median pressure values and thus better-distributed load across the sites (64, 51, 66, and 51 mmHg for the occiput, right mandible, left mandible, and chin, respectively), which may indicate higher conformity of fit. Across all collars, individuals experienced interface pressure values exceeding 100 mmHg, particularly over the occiput, which exceeds the 90 mmHg interface pressure reported to occlude microcirculation (Worsley et al., 2020). If applied for extended periods, even pressures below this threshold can significantly damage the soft tissue (Gefen, 2009a, 2009b). These interface pressure values at the occiput and mandible for the Aspen Vista, Stiffneck Select and Miami J collars recorded are similar to those recorded in the literature (Plaisier et al., 1994; Tescher et al., 2007; Tescher et al., 2016; Worsley et al., 2018). Recorded values here fall in the range between mean and peak values of 26.9 mmHg (mean) and 64.5 mmHg (peak) at the occiput and 24.1 mmHg (mean) and 110.0 mmHg (peak) at the mandible recorded by Tescher et al. (2016). This is as expected, given their study’s higher sensor area and resolution. In several cases, it was observed that the collar did not make complete contact with the chin portion of the collar, resulting in very low or zero pressure. In these cases, the jaw shape meant most contact was carried on the mandibles.

The Stiffneck was the most restrictive compared to the other three collars aligning with a previous study comparing the Stiffneck and Aspen Vista collars (Worsley et al., 2018). There was substantial inter-subject variability in the range of motion for each collar design. This demonstrates a high variance in the risk of motion-related injury to the spine. Additionally, range of motion was correlated with some anthropometric measurements, indicating that patient morphology influences the collar’s efficacy. The range of motion evaluated in this study compares well with those reported in the literature (Tescher et al., 2007; Tescher et al., 2016; Worsley et al., 2018). Though the transverse rotation range is generally higher than in the literature, Tescher et al. (2016) reported transverse rotation to be half that of this study for the Aspen collar. However, Worsley et al. (2018) reported a similar transverse rotation range to the present study. To better meet the needs of all collar users, further investigation is needed into the causes of this high variability. This research area would benefit from an objective measure of the movement restriction.

This study identified significant differences in the relative humidity and temperature at the device-skin interface between the different collar designs. Though limited differences in skin temperature were observed between collars, all designs contributed to an increased skin temperature when compared to normal physiological ranges for a resting adult male (30-34°C) (Gagge et al., 1967; Winslow et al., 1937). The Stiffneck and Philadelphia collars had higher relative humidity than the Aspen Vista and Miami J collars. Black et al. (1998) also found increased skin humidity in a Philadelphia collar compared to an Aspen collar. The padding materials likely contribute significantly to temperature and humidity increases. The Stiffneck and Philadelphia both use an uncovered, closed-cell foam. In contrast, the Aspen Vista and Miami J collars use an open cell foam with a layer of fabric covering the surfaces that contact the skin. Compared to open-cell foam and permeable fabrics, closed-cell foam reduces heat and moisture transfer to the surrounding environment (Glicksman, 1994). Humidity and temperature play an important but not fully understood role in the risk of pressure ulcer development (Kottner et al., 2018). Changes in the microclimate around the skin affect its mechanical properties and ability to withstand extended mechanical loading (Kottner et al., 2018). Animal models demonstrate that temperatures above 35°C significantly impact the viability of loaded soft tissues (Kokate et al., 1995). The elevated temperatures and high relative humidity for specific collar designs observed in this study likely put the underlying soft tissue at increased risk of pressure ulcers. Skin hydration and TEWL are also affected by the device-skin interface. However, this study revealed no changes in TEWL or hydration at the chin following collar application. This could have been due to the limited contact at the chin site for some participants and the relatively short application period of the device. Indeed, other studies with more prolonged device application have demonstrated changes in TEWL and hydration (Abiakam et al., 2023; Kottner et al., 2015).

Patient discomfort has not been fully reported in studies evaluating the efficacy of cervical collars. Discomfort negatively contributes to patients’ experience and quality of life, adding to the already traumatic situation that warrants collar use. This potentially contributes significantly to device abandonment in the rehabilitation setting. Patient-reported discomfort has only been recorded for the Aspen Vista and Stiffneck Select collars in the literature; discomfort for the Aspen Vista is similar to that recorded in the present study, but discomfort associated with the Stiffneck Select is higher than previously reported (Worsley et al., 2018). In an observational study, Ham et al. (2016) found a high incidence of pain (63.2%), occurring most frequently at the occiput and for female patients, but the collar design was not mentioned. Discomfort was correlated with the pressure at the occiput and the relative humidity at the interface. Elevated temperature, in combination with the build-up of moisture, is associated with increased discomfort (Gagge et al., 1967). This study revealed elevated temperatures across all collar designs and significantly elevated interface humidity for specific designs, the compounding effects of which likely contribute to the discomfort levels observed. Further work investigating the dominant factor contributing to perceived discomfort would help focus research on collar designs for improved overall user experience.

## Limitations

The present study recruited young, healthy participants. This limits the generalisability of the findings to the population of collar users, who often have comorbidities. The collars were fitted according to the researcher’s interpretation of the manufacturer’s guidelines and not by a trained orthotist which may also contribute to the high variability in some results. However, collars often need to be donned and doffed by the user or their family members; it was thus relevant to involve lay persons.

Pressure was measured at discrete points at locations specifically susceptible to pressure injury. This does not fully describe the interface pressure distribution nor the mechanical characteristics of the underlying tissue. Additionally, the pressure sensors were limited to reading values below 258 mmHg, which was exceeded in several instances, especially for the Stiffneck collar at the occiput. Therefore, there was a ceiling effect on the observed pressures at this site. The true extent of the mechanical interactions need to be explored with sensors incorporating a wider operating range. Finite element modelling could be used to further estimate interface pressure distribution and the stress and strain within the underlying tissue, whereby potential outcomes could also be verified by the sensor measurements in this study.

The short collar application could have led to the minimal physiological response of the skin. Prolonged donning periods could reveal more about the skin’s response. Additionally, the physiological response measurements were limited to the chin site as this was the most convenient. A high degree of variability in interface pressure was observed between participants at this site, including instances of zero pressure, further limiting interpretation of findings. Future research focussing on the occiput site could also be valuable.

Participant discomfort was measured with an 11-point numerical rating scale consistently across all collar applications. This method has not been validated for the present study setting, limiting the interpretation of results. Despite not being validated, this method has been used previously to successfully identify significant differences in collar comfort (Worsley et al. 2018).

Although the parameters measured in this study are known to influence the risk of pressure ulcers, the exact relationships with pressure ulcer development – especially in combination – are unclear. Additionally, there are many other intrinsic and extrinsic factors that will influence the risk of collar-related pressure ulcers (Coleman et al., 2014). Therefore, the significance of collar design considerations remains uncertain.

# Conclusions

This study investigated several key bioengineering factors with a view to preventing medical device-related pressure ulcers which have been identified by Bader et al. (2019). By exploring a range of factors at the device-skin interface this study has enabled a better understanding of the most significant differences between collars and the influence of collar design and materials on key outputs. The data collected in the present study demonstrate apparent differences in the interface pressure and humidity across common collar designs. A more than twofold increase in pressure at the occiput was observed in the Stiffneck collar compared to the MiamI J collar. Collars that used closed-cell foam padding had 20 %RH higher humidity at the interface compared to those that used lined open-cell foams. Both factors are known to play a significant role in the increased risk of pressure ulcers. This was also evident in perceived discomfort for the end-users.Improving the design and fit of cervical collars will reduce the incidence of collar-related pressure ulcers, improve the patient’s quality of life, and reduce the cost of care.

# Statement of competing interest

The authors declare that they have no conflict of interest.

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Fig. Collar designs; A) Miami J standard collar with sternal pad (Össur UK Ltd, UK), B) Stiffneck Select collar (Laerdal Medical, UK), C) Philadelphia Adjustable Tracheotomy collar (Össur UK Ltd, UK), D) Aspen Vista collar (Aspen Medical Products, USA).

Fig. Test protocol timeline schematic for one collar application.

Fig. Collar application with sensor locations indicated, occiput (square), right and left mandibles (circles), and chin (hexagon).

Fig. Plot of interface pressure for four collars across four measurement locations (Occiput, right mandible, left mandible, and chin), pressure sensor upper limit (258 mmHg) indicated with horizontal line.

Fig. Plot of microclimate parameters (mean (± 1 std) temperature and relative humidity) at the collar-skin interface for each collar at the chin, right mandible, and left mandible.

Fig. Plot of perceived discomfort for each of the collar where 10 in most discomfort and 0 is no discomfort.

Fig. Spearman correlation coefficient between each of the test parameters. Statistical significance is indicated by bold fonts, where p < 0.05.