REPORT

An ECG Based Heart and Breathing Rate Monitor for Use in Space

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Abstract: There is a need to monitor the physiological signals of astronauts, both for their safety and to study the impact of this foreign environment. NASA recently released a Request for Application for a device that can measure heartbeat and respiration to determine whether an astronaut has accidentally fallen asleep. To address this, a device described in the following study was created to determine changes in breathing rate within a 30 sec. interval and to determine changes in heart rate within a 10 sec interval from ECG signal. The device was further designed to be as noninvasive and compact as possible. The device presented here is able to measure heart rate with an uncertainty of 1.62 BPM and breathing rate with an uncertainty of 0.013 Hz, indicating a high level of precision. While this device is able to capture these signals with high accuracy and precision, further improvements can be made with advances in electronic component tolerance, higher order active filtering, and more rigorous device testing.

Introduction:

Astronauts are exposed to a variety of foreign conditions while in space, such as differences in lighting, gravity and radiation. In addition, astronauts are required to perform complicated procedures and experiments in which a mistake could be costly and potentially life-threatening.[2] For this reason, there is a great interest in monitoring the effect of space on physiological signals as well as a great need to monitor these signals for the safety of astronauts.

As a result, NASA recently released a Request for Application for an instrument that can measure heartbeat and respiration in space to determine whether an astronaut has accidently fallen asleep.[1] The following study describes a solution for this Request for Application. The preliminary device created attempts to monitor ECG signals and extract breathing and heart rate such that changes in breathing rate could be detected in a 30 second interval and changes in heart rate could be detected in a 10 second interval. In addition, the solution was held to a few practical constraints such as requiring as few and as noninvasively-placed electrodes as possible and restricting the size of the circuit used. Several systems have already been used or proposed to measure the physiological signals of astronauts. For example, a proposed sensor for use during astronauts' extra-vehicular activities includes a plethysmography sensor, skin resistance sensor, skin temperature sensor and pulse oximetry sensor.[3] Despite these pre-existing devices, we attempt to create a highly accurate device specifically for the measurement of breathing rate and heart rate. The preliminary device mentioned in this paper is able to satisfy the previously mentioned constraints while accurately measuring heart rate and breathing rate as validated by external measurements. However this work represents a preliminary exploration of a monitoring device. Further work must be completed to increase the accuracy of measurements and rigor of device testing.

Results:

Table 1: Analyses performed on ECG data

Subject	Average Heartbeat Duration (s)	SEM Heartbeat Duration (s)	Average BPM (Beats/min)	SEM BPM (Beats/min)	Average QT interval (s)	SEM QT interval (s)
Subject 1	0.74	0.0063	81.49	0.98	0.40	0.0066
Subject 2	0.78	0.0028	77.11	0.27	0.41	0.0052
Subject 3	0.76	0.0050	79.08	14.28	0.47	0.0043
Subject 4	0.84	0.0058	71.08	9.99	0.49	0.0021

Digital ECG data, obtained from the Biopac (MP35, Biopac, Goleta, CA) ECG filtering setting, was first analyzed to obtain a method for quantifying the wave. Algorithms were developed to measure average heartbeat duration, average beats per minute (BPM), and QT interval (Table 1). Notice that ECG is able to pick up heartbeat changes between

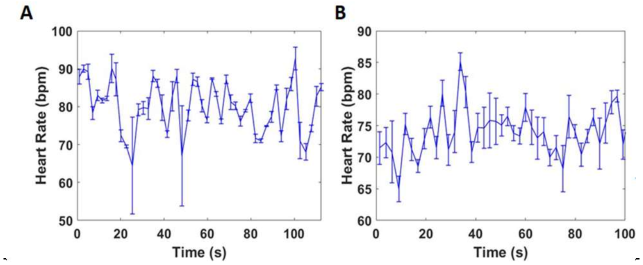

Figure 1: Shown are representative plots of the instantaneous heart rate for **(A)** Subject 1 **(B)** Subject 2. Data was averaged across three consecutive time points (N=3) and is shown as mean plus/minus standard error. Here, for clarity, the first approximately 100 seconds of the recording is shown. Note: plots for subjects 3 and 4 are provided in the supplemental information

test subjects along with changes in QT interval. We found that the value of the QT interval was within the range of normal values, which is 0.38- 0.52 seconds, for all subjects.[6] The typical orientation for the QRS vector is between -10 to 110 degrees.[7] Instantaneous heart rate was also computed for all subjects (Fig 1). This allowed for real time estimation of the heart rate and was tagged with an uncertainty of the time period that the estimate occurred over.

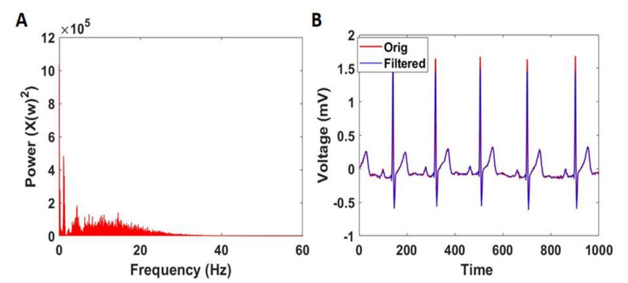

Figure 2: Shown are elements of the frequency domain analysis performed on the ecg signal. **(A)** Plotted is a power spectrum of the signal obtained from taking the discrete Fourier transform of the signal. Notice the high power elements at frequencies of approximately 0 and 1.2 Hz. **(B)** Shown is the original signal (orig) along with the signal after implementing a low pass filter of 20 Hz and a high pass filter of 0.1 Hz (Filtered). Notice that the filtered signal is smoother and takes a more predictable pattern without sacrificing significant information from the original signal.

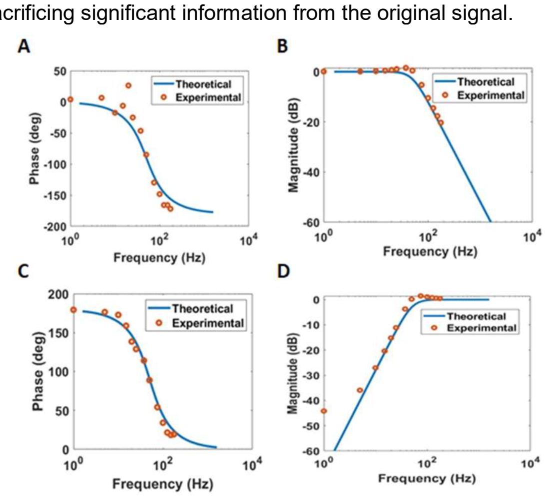

Fig. 3. Shown are theoretical and experimental phase responses for **(A)** low pass active filter and **(C)** high pass active filter. Additionally magnitude response for **(B)** low pass active filter and **(D)** high pass active filter. These plots show high overlap of experimental and theoretical results

Frequency domain analysis was then performed on the highest quality ECG signal. The mean of the signal was first subtracted from the signal. The Discrete Fourier Transform was taken to visualize the signal's power spectrum (Fig 2A). A majority of the power lies around frequencies of approximately 0 to 20 Hz. There is a strong power peak at approximately 1.2 Hz. A spike at 60 Hz was not observed. However, if this spike was visualized, it would be associated with noise from the wall power socket. The essential bandwidth of the signal was found to be 26.5 Hz. From a physiological lens, the peak at 1.2 Hz likely corresponds to the heart rate of the subject. There are also peaks in the range of 0 to 0.3 Hz. Physiologically, these peaks likely correspond to the breathing rate and subject movement. In addition, there are higher frequency elements extending to around 20 Hz. These peaks represent the higher frequency components of the QRS complex. Digital filters were then designed in MATLAB to filter the ECG signal. A low pass filter of cutoff 20 Hz was applied along with a high pass filter of cutoff 0.1 Hz. The amount of power remaining after filtering was found to be 74.9%. This resulted in a smoother signal that did not sacrifice features of the original signal (Fig 2B). The analyses shown in Table 1 were performed again after filtering. Filtering did not improve the measurement of heart rate or duration of heart beat. However, the standard error of the measurement of the QT interval did increase. These results are in line with the observation that filtering makes more significant changes to the baseline of the wave rather than the peaks.

Prior to designing the device, UAF42 active filters and RC passive filters were explored. Bode plots considering the magnitude and phase response of second order low pass and high pass Butterworth UAF42 active filters of cutoff frequency 50 Hz were produced (Fig. 3). Here experimental data, found by using an oscilloscope to measure input and output signals to each filter, was compared with theoretical transfer functions computed in MATLAB. The differences between theoretical and experimental data give a good gauge as to how much the difference between values specified in Filter42 program and values actually used to construct the filter contributed to error. This was quantified by computing the RMSE between the experimental and theoretical data. For the low pass filter, an RMSE of

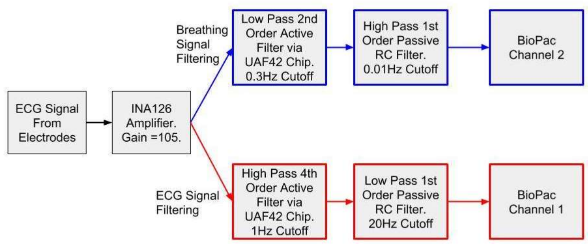

Fig. 4. Diagram showing signal filtering scheme used for design challenge. Within the breathing signal filtering branch, a high pass 1st order passive RC filter with a 0.01Hz cutoff was used to eliminate DC noise. Within the ECG signal filtering branch, a low pass 1st order passive RC filter with a 20 Hz cutoff was used to eliminate high frequency noise from sources such as the line voltage.

1.8467 dB was found for the magnitude response and 22.1664 degrees was found for the phase response. For the high pass filter, an RMSE of 2.0768 dB was found for the magnitude response and 7.1037 degrees was found for the phase response. In both the theoretical and experimental cases we observed a DC gain of 1. This is the expected result, given that we did not design the filters to have a gain. In both the high and low pass filters the observed cutoff of approximately 50 Hz matched the expected cutoff of 50 Hz. Thus, these experimental values match the theoretical predictions. In addition to building active filters, a 50 Hz high pass RC filter was constructed. Bode plots of this filter were also produced (Supplementary Figure. 2). The results showed that RC filters yielded a much more gradual frequency cutoff (gain drop off). A more detailed comparison is given in methods.

After examining the frequency domain representation of the ECG signal, a plan for a device to separate breathing rate (12-18 breaths/min) from heart rate (60-100 beats/min) was developed (Fig 4). The electrical setup of the device consisted of instrumental amplification (gain of 100) followed by filtering into breathing and ECG channels. Based on the previous frequency domain analysis (Fig 2A) specific cutoff frequencies were used (Fig. 4). The accuracy of the filtering design created was verified by producing bode plots of the two filtering channels (data not shown).

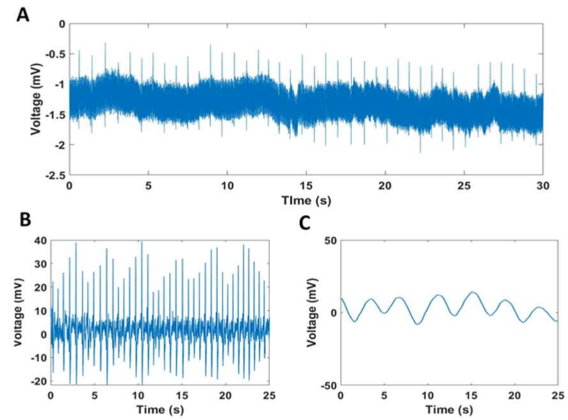

Fig. 5. Shown are representative plots from ECG processing using our custom filtering device. **(A)** Raw input data read in from the ecg electrodes. **(B)** Filtered ECG signal. **(C)** Filtered breathing signal. Note that, in addition to filtering, the initial raw signal was amplified. The breathing rate **(B)** takes the form of a slow sinusoidal wave, while the ECG signal **(A)** resembles the previously seen ECG signals

Device performance was evaluated through two lenses: precision and accuracy. The device was also tested across three subjects to ensure functioning on a range of baseline heart/respiratory rates and activity. The device was additionally tested to ensure that the system was able to detect changes in heart rate that occur over a 10 second period and changes in respiration that occur over a 30 second period (per design challenge specifications). Representative output from the filtering device shows a low amplitude noisy signal, being filtered into two channels: one for breathing activity and one for heart activity (Fig 5). Notice the clear peaks in these filtered channels that allowed for the calculation of heart rate and respiratory rate (Fig 5B, C).

To gauge the precision (spread of device measurements for the same task) we used the device to measure resting heart rate and constant breathing rate across three test subjects (Fig 6A, B). To quantify precision of the heart rate measurements, each subject's resting heart rate over 15 second intervals was taken 5 times with the device (Fig 6A). It was found that the device presented a standard deviation of 3.16 BPM, 3.25, and 4.48 BPM for Subject 1, Subject 2, and Subject 3 respectively. These standard deviations show a very small amount of variance, indicating high precision of the device in measuring heart rate. To report a parameterized error associated with overall device measurement of heart rate, the average of the SEMs for Subjects 1, 2, and 3 were taken. From this, an overall error of 1.62 BPM (SEM) was found to be associated with the device's measurement of heart rate. Similarly, to quantify precision of the breathing rate measurement, each subject's breathing rate was measured 5 times over 30 second intervals (Fig 6B). Subjects were ensured a constant breathing rate of once every four seconds (0.25 Hz). It was found that the device presented a standard deviation of 0.031, 0.032, and 0.023 Hz for Subject 1, Subject 2, and Subject 3 respectively. Again, this represents a very small amount of variance, indicating high precision of the device in measuring breathing rate. The overall parameterized error associated with device measurement of breathing rate, was calculated by taking the average SEM for subjects 1,2, and 3. From this, an overall error of 0.013 Hz (SEM) was found to be associated with the device's measurement of breathing. This task can also be used to gauge the accuracy of the device (closeness of the measured value to the known value). 95% confidence intervals of measured breathing rate for Subject 1, Subject 2, and Subject 3 respectively were as follows: [0.23,0.33], [0.20,0.29], [0.22,0.29] Hz. Given that 0.25 Hz is contained in the above three confidence intervals, there is no evidence at a significance level of 0.05 that the measured values for breathing rate is different than the known rate of 0.25 Hz.

To further gauge the accuracy of the device and whether the system was able to detect changes in heart rate that occur over a 10 second period and changes in respiration that occur over a 30 second period, additional experiments were performed. Test subjects worked their heart rate up to approximately 120 BPM and then allowed their heart rates to fall back towards resting for a minute. The change in heart rate every 10 seconds was measured manually by counting and by the device (Fig 6C). The mean and SEM were calculated for both the experimental and known data across three trials per subject. Paired t-tests were run across the three subjects under the null hypothesis that the device measurement was the same as the known value. At a significance level of 0.05, there was no evidence of differences between the two data, indicating that the device remained accurate across the three test subjects and was able to pick up changes in HR occurring over 10 seconds. Similarly, test subjects were instructed to change their breathing rate from 0.33 Hz to 0.2 Hz to 0.125 Hz over 30 second intervals. The known breathing rate is plotted alongside the mean device measurement (N=3) with standard error for these 30 second intervals. T-tests under the null hypothesis that the mean device measurement was equal to the known breathing rate returned no evidence of any difference, at $\alpha = 0.05$. The device is also able to accurately pick up changes in breathing rate over 30 second intervals.

Discussion:

This study demonstrated the creation of an electrical filtering device that was able to successfully monitor ECG signals and extract breathing and heart rate. Using this device, changes in breathing rate could be

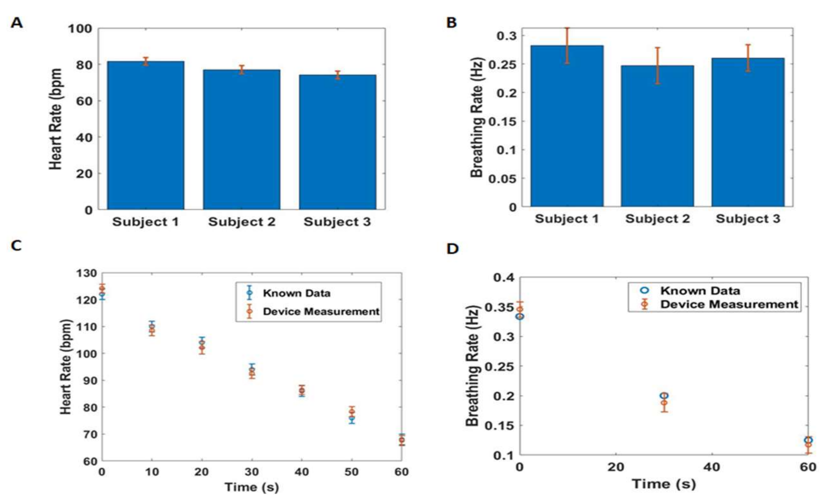

Figure 6: Shown are the results of the experiments run to validate the device. **(A)** Shown is the device's measurement of baseline Heart Rate over 15 second intervals (N=5) for 3 different test subjects. This is plotted as mean plus/minus standard deviation. The small standard deviations of measurement indicate high precision of the device. **(B)** Shown is the device's measurement of Breathing Rate (subjects were instructed to breath once every 4 seconds) over 30 second intervals (N=5) for 3 different test subjects. This is plotted as mean plus/minus standard deviation. Notice the higher standard deviations for these measurements. **(C)** Shown is the device's response to a changing heart rate every 10 seconds. Results are plotted for a representative subject (Subject 2, N=3) and are plotted as mean plus/minus SEM for both experimental and theoretical. **(D)** Shown is the device's response to a changing breathing rate every 30 seconds. Results are plotted for a representative subject (Subject 1, N=3) and are plotted as mean plus/minus SEM for the experimental data.

detected in a 30 second interval and changes in heart rate could be detected in a 10 second interval. The device was able to report heart rate with an uncertainty of 1.62 BPM and breathing rate with an uncertainty of 0.013 Hz. This represents a very precise measurement. The features of the design itself are also robust in ensuring a minimum number of electrodes, and a compact electrical circuit. Rather than using multiple leads to measure ECG signal, the device relies on one lead, consisting of three electrodes: the negative electrode on the right arm, positive electrode on the left arm and ground electrode on the bony part of the lower left leg. This allows for astronaut mobility and flexibility, which is important for day to day tasks in space.

As stated in results, before designing the device a variety of passive and active Butterworth UAF42 chip filters were compared. There are a few differences between these two types of filters. One disadvantage of the UAF42 filters is that they require a power source, whereas passive filters operate only through signal input. One strength is that the UAF42 chip has the ability to introduce gain while passive filters do not without an additionally circuit element. Additionally, a strong advantage of the UAF42 chip is the ability to have a sharper frequency cutoff compared to a passive filter. The steepness of this frequency cutoff increases with order.

When creating the device several important factors were considered. Overall, the device was kept compact and attempted to minimize the number of components while maximizing accuracy of the measurements The normal respiratory rate for an adult is 12-18 breaths/min (0.2-0.3 Hz) and the normal heart rate is 60-100 beats/min (1-1.67 Hz).[1] Given the small frequency difference between breathing and heart rate, we leveraged active components' ability to have sharper cutoff frequencies to separate these two signals. A second order UAF42 active Butterworth low pass filter with a 0.3 Hz cutoff was implemented in order to isolate this breathing signal, and a fourth order UAF42 active Butterworth high pass filter with a 1Hz cutoff was implemented to isolate heart rate.[1] Here, a fourth order filter was used at the interface of breathing and heart rate to minimize the amount of overlap between channels, due to its steep frequency cutoff. Two additional passive filters were used to further clean the data. A passive RC first order high pass filter with a 0.01 Hz cutoff was placed after the UAF42 active low pass filter in order to eliminate DC component noise at zero frequency present in the breathing channel. A passive RC first order low pass filter with a 20 Hz cutoff was placed after the UAF42 active high pass filter to eliminate noise generated by line voltage and other high frequency artifacts. Refer to Figure 3 for a diagram of the filtering scheme used for the design challenge. The basis for including these passive elements was the previous digital analysis performed on ECG signal. From this analysis, it was known that DC noise and line voltage presented key noise generators.

Error in this study, quantified by the SEM of the device's measurement of heart and breathing rate, stemmed from a few key areas. First, while the effect was limited, the device was still affected by patient movement. Movement by the patient caused artifacts to appear in the ECG and breathing waves that impacted the calculation of heart and breathing rate. Additionally, resistors used in this study had 5% tolerance. This impacts the ability to build precise filters that depend on an exact resistance value, and as such future studies could use resistors with a lower tolerance. The filtering device presented in this work can also be improved by replacing all passive filters with active filters in order to increase the slope of the frequency cutoffs. The order of all filters can also be increased in order to increase the rate of gain drop off. Increasing the rate of gain drop off will help to eliminate unwanted frequencies permeating into the final signal. Lastly, more rigorous device testing should be performed. Rather than compute known heart rate measurements by simply counting, more advanced technology that is known to produce accurate results should be used. Thus, future studies should build on the efficacy of this device by testing the device against accurate known heart rate measurements such as those produced from an Apple Watch or commercial heart rate monitor. Additionally, future studies should investigate innovative methods to minimize the effect of patient movement on device function. Lastly, future studies may utilize more expensive, accurate electrical equipment to create devices with stronger filter accuracy. However, the implications of this study are large. We have successfully constructed a device that astronauts may wear to monitor heart and breathing rates and detect changes within 10 and 30 seconds respectively. This device fits into the broader active field of physiological processing and may build on the work of others such as the creation of a contact-free system to measure heart rate, respiration rate, and body movements during sleep.[5] Furthermore, future work could be done in order to use a device like the one presented to diagnose the severity of medical conditions in which ECG is affected such as Wolff-Parkinson-White Syndrome. Ina condition wherein during exercise testing the typically seen delta wave of a Wolf-Parkinson-White patient's ECG can disappear, indicating that this patient is at a lower risk of death due to sudden arrhythmia.[8]

Materials and Methods:

Measuring ECG Data. ECG data was gathered using a 3-electrode setup, with the negative electrode on the right arm, positive electrode on the left arm and ground electrode on the bony part of the left bottom leg. This setup was chosen because it exhibited the least noise and drift and had the sharpest peaks. In addition, it showed the least variation due to subject movement, which is important for use with astronauts. Data was collected using the under the ECG (0.05-150 Hz) mode of the Biopac Data acquisition system (MP35, Biopac, Goleta, CA).

Digital ECG Analysis: The gathered ECG data was analyzed in order to extract components, such as the duration of a heartbeat, average and instantaneous heart rate, and the average QT interval. Detailed descriptions of the algorithms are given along with the code in the supplemental information. Frequency domain analysis was also performed on the data set. ECG signal from Subject 1 was used for this analysis as it was the highest quality signal. Again, MATLAB was used to take the raw ECG signal and subtract the DC (mean) component. The magnitude and power spectrum of the discrete fourier transform performed was also computed through MATLAB code.

Experimental and Theoretical Implementation of Digital Filters: MATLAB was used to theoretically design the filters that were to be used throughout our experiments. Passive and active low pass and high pass filters were modeled, producing both phase and magnitude response plots for the filters. This was accomplished by first finding the transfer function of the filter used, and then making use of MATLAB's tf and bode commands to generate the corresponding bode plot. Transfer functions for UAF42 active filters used were determined via the UAF42 datasheet. Experimentally, passive and active filters were created and tested using a myDaq Student Data Acquisition Device (myDAQ University Kit, National Instruments, Austin, TX). The function generator feature of the myDAQ provided the power signal to the filter, and the oscilloscope was used to compare the output of the filter to the original input signal. With this setup, an RC high-pass filter with cutoff frequency of 50Hz was designed and an active 2 pole low-pass filter with cutoff frequency of 50Hz and DC gain of 0 dB. The resistor and capacitor values for these two passive RC filters were chosen based off of the following equation: Cutoff Frequency=1/(2**R*C).[4] Theoretical gain and phase responses were modeled using the method outlined above, which were then compared to experimental data.

Device Construction: In order to successfully design a filter with the capability of functioning as a dual heart rate/respiration monitor that has the ability to separate breathing from heart rate signal, we designed a multi-component filter device that leverages both passive and active filters. To design this filter signaling scheme, the input ECG data was first amplified through the use of a INA126 amplifier (INA26 Micropower Instrumentation Amplifier, Texas Instruments, Dallas, TX). After this point, our filter design was split into two separate branches, one to isolate breathing and the other to isolate the heart rate. For the heart rate components of this filter, a high pass 4th order UAF42 Chip active filter with cutoff frequency of 1Hz was combined with a passive low pass 1st order RC filter with frequency cutoff of 20Hz. To design the 4th order active filter, the Filter42 program was used. To filter the breathing signal, a low pass 2nd order active filter UAF42 chip with 0.3Hz cutoff frequency was used in combination with a high pass 1st order passive RC filter with a 0.01Hz cutoff frequency. The resistance and capacitance values used along with a diagram of the entire device circuit is shown in the supplemental section Figure 3 (circuit diagram created using circuitlab.com).

Device Validation: To validate the device design we conducted a series of experiments to compare known breathing or heart rates to device measurements. Across all experiments, known heart rate values were defined by the user counting his jugular pulse. Known breathing rate was defined as the user breathing at a given rate, for example once every three seconds. Statistical uncertainty was reported as standard error of the mean and device variation was reported as standard deviation. When comparing known and recorded datasets t-tests were used at a significance level of 0.05.

Conclusions:

A device designed to monitor the breathing and heart function of astronauts in space was developed and presented in this study. The device performed with a high level of precision and accuracy, capturing heart rate with an uncertainty of 1.62BPM and breathing rate with an uncertainty of 0.013Hz. This device was able to measure changes in breathing rate in an interval as short as 30 seconds and changes in heart rate in an interval as short as 10 seconds. Future studies can improve on the work presented in this study by creating devices with resistors of lower tolerance and strictly active filtering components of higher orders to increase frequency cutoff (gain drop off). Additionally future studies should attempt to create a device that is less sensitive to patient movement.

Supplementary Information (SI):
The supplementary information submitted with this report includes plots of instantaneous heart rate for subjects 3 and 4, magnitude and phase plots containing theoretical and experimental data for a passive high pass RC filter built for testing purposes, a circuit diagram of the device created to monitor astronaut's heart and breathing rate, and MATLAB source code used for data analysis.

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