A Microcontroller-Based Portable Electrocardiograph Recorder

José J. Segura-Juárez, David Cuesta-Frau*, Luis Samblas-Pena, and Mateo Aboy

***Abstract*—We describe a low cost portable Holter design that can be implemented with off-the-shelf components. The recorder is battery powered and includes a graphical display and keyboard. The recorder is capable of acquiring up to 48 hours of continuous electrocardiogram data at a sample rate of up to 250 Hz.**

***Index Terms*—Biomedical signal acquisition, data acquisition system, electrocardiogram (ECG), Holter recorder, microcontroller.**

I. Introduction

Over the last two decades portable physiologic signal acquisition systems have been developed that are light, small, and capable of recording multiple signals for up to 48 hours. These systems are used in electrocardiography (ECG) studies to detect infrequent cardiac arrhythmias or transitory cardiac function abnormalities often related to the tensions of daily life. These are often called Holter recorders, named after an ECG recorder developed by N. J. Holter in 1961 [1]. The first Holter recorders utilized a tape to store the signals, since large capacity solid state memories were not available yet. Today, the signals are recorded in memory cards that can be easily transferred to a computer workstation for analysis.

We describe a complete design of a Holter recorder. The design uses off-the-shelf components, including an embedded microcontroller, and can be easily built with inexpensive components. This recorder can also be used as a reference model to build portable acquisition systems for other biosignals.

The computational power of a simple microcontroller is sufficient for a Holter recorder, since the most complex and time consuming tasks are usually carried out offline, in a personal computer. Sampling frequency may range from 100 Hz to 1 KHz, depending on the ECG analysis to be performed. For long term acquisition, a value around 250 Hz is sufficient.

Several other ECG acquisition systems designs have been described [2]–[4], but none of them achieve the performance of commercial Holter recorders regarding the portability and nonvolatile data storage.

II. Recorder Description

This section describes the main components of the recorder. Fig. 1 shows a block diagram of the system and how the components are interrelated.

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A. Electrodes

There are many different electrodes types commercially available [5]. They should have an adhesive area to fix them properly to the skin, clip-on wires, and a conductive gel, for stable low-noise recording. Active electrodes can also be used to improve the resolution of the recording [6].

Our Holter recorder uses seven electrodes (three lead configuration) that may be placed at locations suited to the application. Three pairs of the electrodes are differential voltage inputs and one serves as a reference [7]. This reference electrode and its associated circuit (Figs. 2 and 3) offer a large reduction of common mode voltage magnitude by actively reducing the voltage difference between patient and the ECG amplifier common by means of the so-called driven-right-leg circuit design [8]. This connection is electrically safe for the patient.

B. Cable

It consists of seven lead wires that are utilized to create a three-channel ECG recording. These wires are shielded and twined in order to reduce the induced electrical noise. In the ECG amplifier end, a locking detent system is desirable to prevent unwanted or accidental disconnection and provide solid wire-to-contact connection. Many standard connector accessories that meet these specifications are commercially available [9]. The recorder includes a test connector to ensure the patient hook-up is being done properly.

The cable shield serves as the ground wire. It is connected to the common mode voltage of the integrated precision instrumentation amplifier (described later) through a voltage follower, which diverts interference currents induced in each wire toward ground. This connection is used to reduce the effect of the parasitic capacitances that appear between the wires and ground and, therefore, avoids input impedance reduction. The equivalent circuit of the shield is shown in Fig. 2.

C. ECG Amplifier

The ECG amplifier block in Fig. 1 consists of a differential amplification stage for each lead. Fig. 3 shows the circuit diagram for this differential amplification stage. It has a common mode rejection of approximately 120 dB. It also has small bias currents and offset voltages to prevent saturation of both the ECG amplifier and analog-to-digital converter (ADC). V_{cc} corresponds to the voltage achieved with the two batteries and the step-up dc–dc converter, as described in Section II-H. $V_{cc}/2$ is considered zero voltage level.

The passive circuit preceding the precision integrated instrumentation amplifier (U_1 in Fig. 3) is an ac coupling that highpass filters the voltage difference [10]. This reduces baseline wander and ECG amplifier saturation. This ac coupling circuit has the following transfer function (taking $R_1 = R_2$):

$$G_{DD}(jw) = \frac{jwR_2C_1}{1 + jwR_2C_1}. \tag{1}$$

This ac-coupling network provides a bias path for the next circuit component without any connection to ground. It allows closed-loop control of the dc common mode voltage by means of a driven-right-leg circuit, which is essential in this application where we have a single power supply ECG amplifier. No resistor can be spared because of the bias path they provide [10]. We chose R_2 and C_1 to obtain a cutoff frequency of 0.01 Hz.

Following an integrated precision instrumentation amplifier, U_1, is found. The general specifications for this circuit component are:

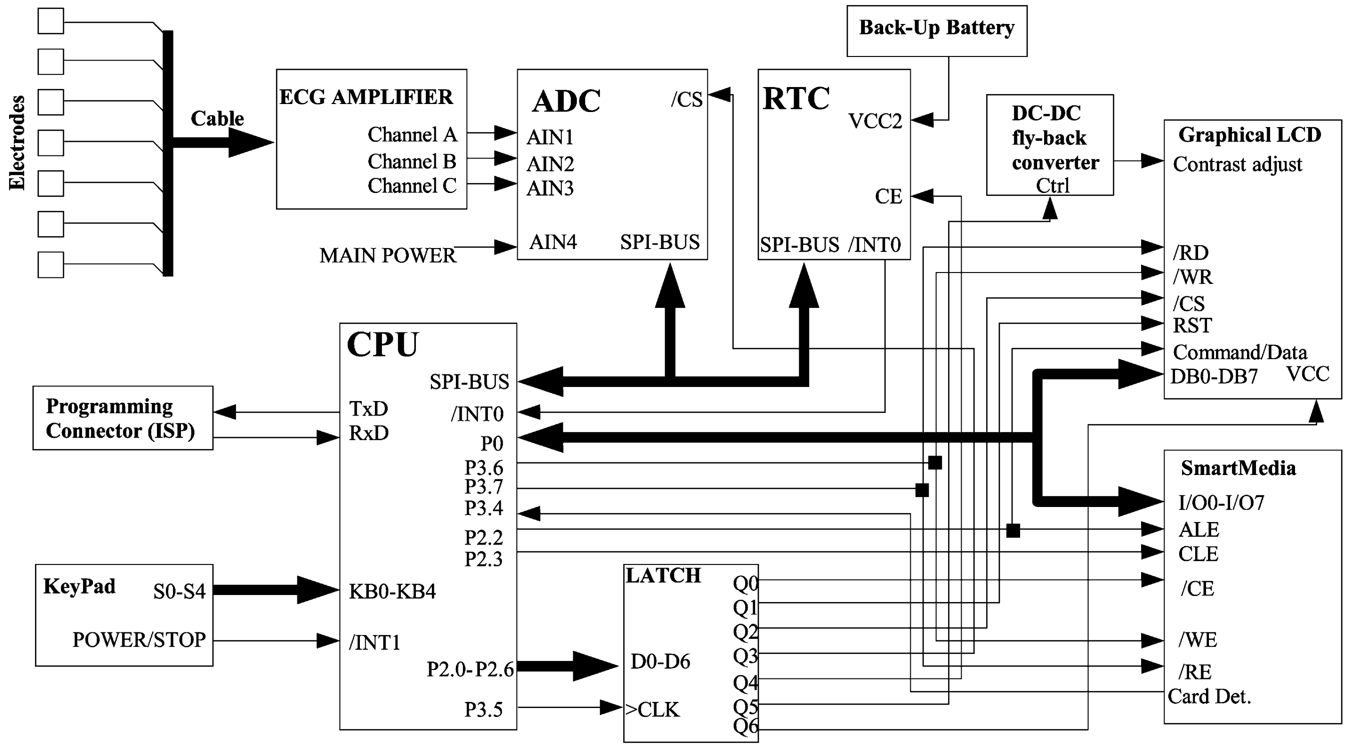

Fig. 1. General block diagram showing the architecture of this Holter recorder. CPU is an Atmel T89C51RC2 microcontroller. Main memory system is implemented using a SmartMedia memory card. Analog-to-digital converter is implemented using the AD7888 circuit.

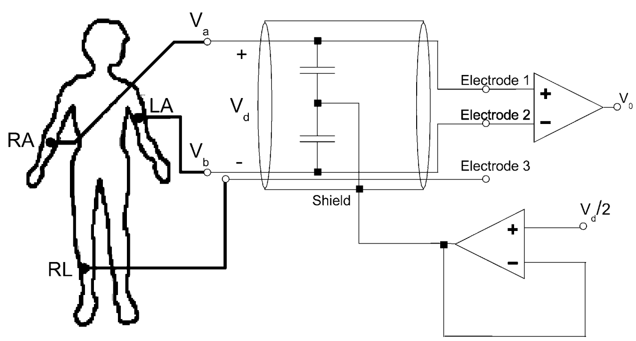

Fig. 2. Equivalent circuit of the shield. Effect of the parasitic capacitances that appear between the wires and ground (shield) should be reduced. Lead electrodes are attached to right arm (RA) and left arm (LA) for signal acquisition, and reference electrode is attached to right leg (RL). Electrode 3 connection is shown in Fig. 3.

- wide supply range: 2 V to 10 V at least;
- high CMRR: >60 dB;
- low offset voltage;
- low bias and offset currents;
- adjustable differential gain G;
- suitable for battery operated systems.

We chose the Burr–Brown INA118U integrated instrumentation amplifier that meets those specifications [11]. External resistor R_G sets the gain, $G = 1 + (50\,\mathrm{k}\Omega/R_G)$. The voltage applied to Ref terminal, V_{Ref}, is summed at the output, where $V_{\mathrm{Ref}}(jw) = -V_{01}(jw)(1/jwR_3C_2)$. This feedback through V_{Ref} performs a second ac-coupling stage whose cutoff frequency is given by $f_{-3\ \mathrm{dB}} = (1/2\pi R_3 C_2)$ [11]. Although the input impedance of the INA118U is very high, a path for the bias currents must be provided, as will be described.

The common mode voltage of the recorder is obtained from U_1, and the cable shield is connected through resistors, labeled as $(R_G/2)$, to it, to reduce the effect of parasitic capacitances. The output voltage of this circuit component, $V_{01}(jw)$, is given by

$$V_{01}(jw) = \left(\frac{jwR_2C_1}{1 + jwR_2C_1}\right)\left(\frac{jwR_3C_2}{1 + jwR_3C_2}\right) GV_d(jw) \qquad (2)$$

where $V_d(jw)$ is the differential input.

The output voltage of the instrumentation amplifier U_1 is filtered by a second-order Bessel low-pass filter with a cutoff frequency of 160 Hz. This is used to reduce high-frequency noise and its bandwidth is sufficient for adult electrocardiograms. A higher bandwidth is necessary for pediatric applications [12]. The transfer function of this filter is

$$G_f(jw) = \frac{1 + \frac{R_B}{R_A}}{1 - w^2\alpha\beta + jw\left[\frac{R_B}{R_A}\alpha + R_4C_4 + \beta\right]} \qquad (3)$$

where $\alpha = R_4C_3$ and $\beta = R_5C_4$.

Thus, the expression of the ECG amplifier output voltage is

$$V_0(jw) = V_{01}(jw)G_f(jw). \qquad (4)$$

U_4 is an inverting amplifier attached to the ground electrode that, in combination with U_3, implements a driven-right-leg circuit [8]. It also provides a return way for the bias currents of circuit U_1.

Operational amplifiers of the circuit are Burr–Brown OPA2336 operational amplifiers [13], which meet the following general specifications:

- low current consumption;
- low offset voltage;
- low offset and bias currents;
- power source level can be 3.3 V;
- high bandwidth;
- high differential and common mode impedances.

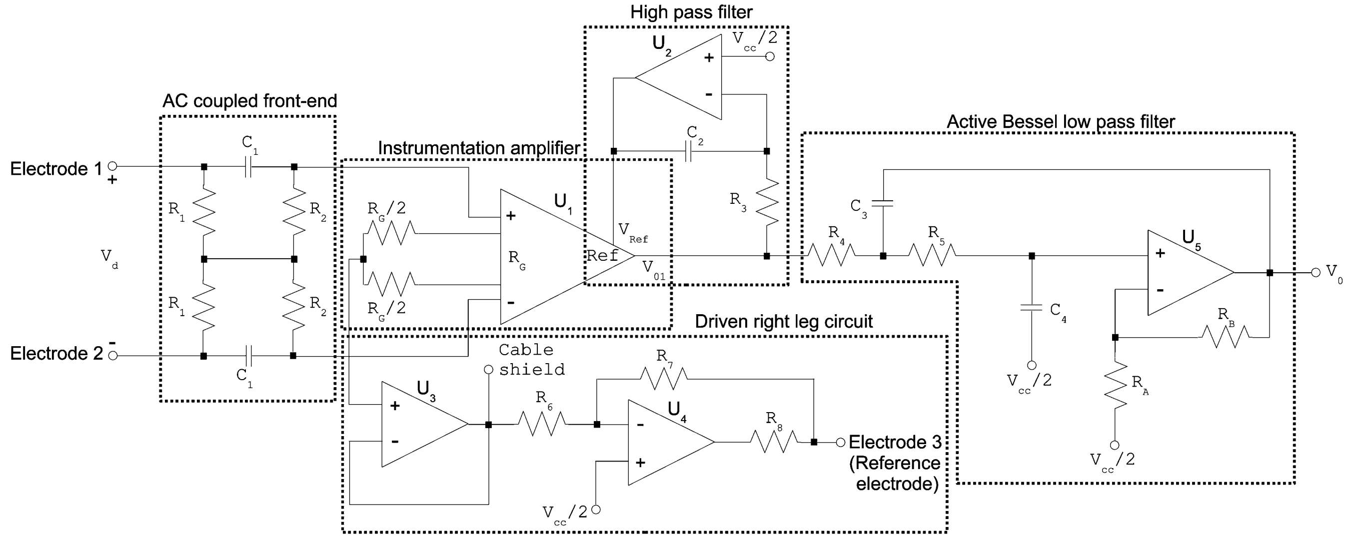

Fig. 3. ECG amplifier of the recorder. Core is the instrumentation amplifier INA118, U_1, supported by some operational amplifiers in different configurations and an input network.

D. ADC

Once the ECG signal has been amplified and filtered, it must be digitized. In order to achieve a proper conversion, some general specifications must be accomplished by the ADC:

- power source level flexibility and low power consumption, in order to work in a battery powered system. It must admit a 3.3-V power source and power saving modes.
- minimal resolution of 8 bits, recommended 12 bits;
- microprocessor simple interfacing;
- appropriate sampling rate for the signal under analysis, 200 Hz at least;
- able to digitize multiple signals. Multiple analog input channels, at least four.

The ADC AD7888 [14] fulfills these requirements. This converter has 12 bits of resolution and 8 single-ended analog input channels. It is capable of a 125-KSPS throughput rate and has low-power dissipation, typically 2 mW for normal operation and 3 μW in power-down mode. The dc accuracy of the AD7888 is typically 1 LSB of integral nonlinearity, $-1+1.5$ LSB differential nonlinearity and an offset error of ± 6 LSB. It includes a serial peripheral interface (SPI).

Three of the converter input channels are used to digitize the corresponding leads, and another one is used to measure the charge level of the batteries. The rest remain unconnected. Communication between the CPU and the ADC is performed through the SPI port.

E. CPU

It consists of a microcontroller that stores data from the ADC, controls the calendar, controls the power failure protection circuit (described later), and controls user interaction through an LCD display and keyboard. To carry out all these tasks, the general specifications of the microcontroller are: multiple parallel ports for peripheral interfacing, timing capabilities, and power saving modes. We chose a microcontroller that meets these specifications, the T89C51RC2 ATMEL microcontroller [16], which is compatible with the well-known Intel family of microcontrollers MCS51.

The parallel ports of the microcontroller are used as the peripherals data buses and for additional control lines (Fig. 1). The LCD and the SmartMedia card are memory mapped. Microcontroller parallel ports 0 and 2 are used as the LCD and the SmartMedia card data and addresses bus. Parallel port 1 is used for communications through the integrated

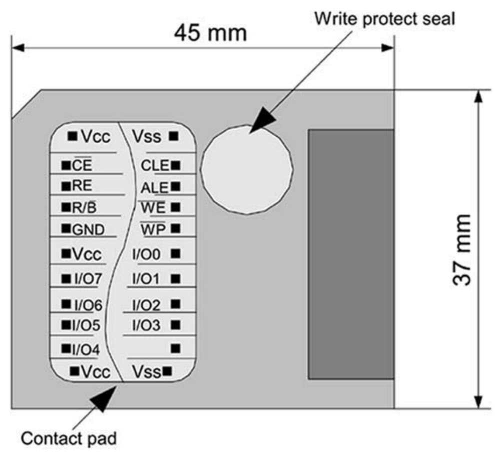

Fig. 4. Schematic diagram of the memory card. SmartMedia memory card is used in this recorder. Data is transferred to the personal computer by means of a special adapter that makes the card work as a floppy disk after suitable formatting.

Fig. 5. Image of a Holter recorder built according to the description in the paper.

SPI interface, which can also be put in shutdown mode to reduce the power consumption. Parallel port 3 bits are used for different control

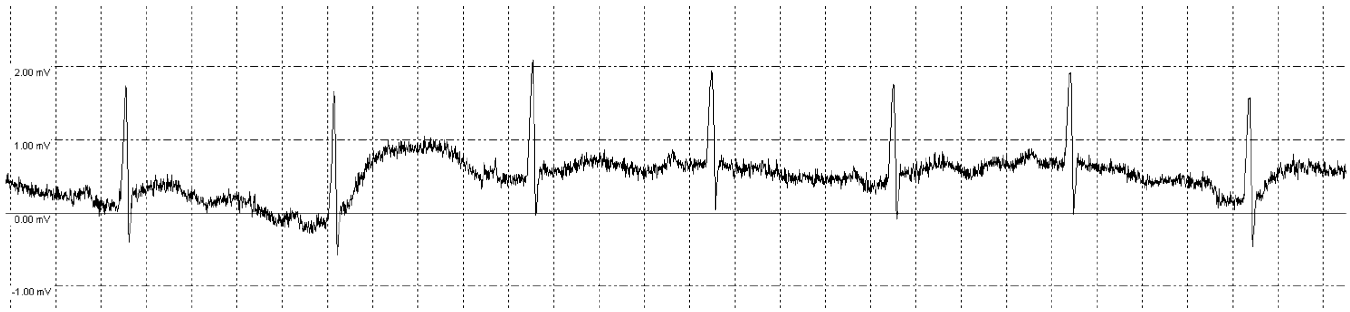

Fig. 6. An example of a real ECG obtained with the recorder described in this paper. No preprocessing has been applied yet. Only a few beats are shown in order to perceive the signal features. Each vertical division corresponds to 0.25 s.

tasks: write and read signals of the memory mapped devices, external interrupts, and additional peripherals control.

Memory error checking is the largest computational load on the microcontroller. This process requires approximately 2 s for each data block of 16 KB. In order to obtain a uniform sampling rate, data acquisition is embedded in this error checking algorithm.

Regarding the memory card, the control is carried out on a hardware level by means of the ports, as mentioned before. The software to control this kind of card checks its validity, its capacity, whether it is formatted or not, the free memory available, and writes the data files.

F. Storage

Long-term portable data acquisition requires a reliable high-capacity storage and low-power consumption memory system. For practical reasons, such a memory system should also be light, small, and cheap.

A device that meets those specifications is a memory card. We chose a SmartMedia model [15] whose format and operation is regulated by the SSFDC [17], [18]. This card provides nonvolatile digital storage of ECG data. SmartMedia cards capacity ranges from 2 to 128 MB. They can be powered using two voltages: 5 and 3.3 V. In Fig. 4, the schematic of this card is depicted.

After the recording, data are downloaded to a computer for further analysis. To serve this purpose, an electronic adapter shaped like a standard 3.5-in floppy diskette is used. The SmartMedia card fits into it through a side slot and works from a 3.5-in 1.44-MB floppy disk drive, even though the card capacity is much greater than those 1.44 MB.

Data are stored using a standard FAT16 logical format since after the recording a floppy disk drive is used to read these data. The ADC takes place during the error checking and parameter calculation of each memory sector: Boot ID, start head, start sector, start cylinder, system ID, end head, end sector, end cylinder, start logical sector, and partition size [17]. Each sample converted is stored in the corresponding memory area, which, after completing a whole data page, will be rewritten in memory using the FAT16 format. Patient can operate a button to mark special events that are recorded along with the signal data.

G. Visualization

The graphical LCD display is a memory mapped peripheral that permits menu-driven user interaction, warning messages, and real-time signal display. The display that we used (PG12864-O) has a 128×64 dot matrix in a 65.5×52.5-mm screen.

The LCD controller requires 3.3 V and the display requires 10.5 V. To obtain this higher voltage, we used a dc–dc MAX749 fly-back converter to obtain -7.2 V [19]. When this is combined with the 3.3-V supply, it is sufficient to power the display. The display draws 2.5 mA when it is on and 15 μA when it is off.

TABLE I
COMPARISON TABLE BETWEEN A COMMERCIAL RECORDER AND THE ONE DESCRIBED

Specification	Philips:ZDP	Our Recorder
Input Impedance	2 M min	100 M
CMRR	>60 dB	120 dB
Recording time (all channels)	24 h or 48 h	0 h to 48 h
Sampling rate	175 Hz	50 Hz to 250 Hz
Resolution	12 bit/10 bit	12 bit
Freq.Response	0.05 Hz-60 Hz	0.01 Hz-125 Hz
Compression	None	None
Memory Type	Flash	Flash Memory Card
Interface	USB	Floppy drive
Visualization	LCD	LCD
Status	LCD	LCD
Battery	AA IEC-LR6	AA (Rechargeable)
Battery life	>48 h	>48 h

H. Power Source

The recorder is powered by two AA batteries in series that produce a nominal supply voltage of 3.0 V. The MAX856 IC [20] is a step-up dc–dc converter that we use to boost the nominal 3.0-V supply to a regulated 3.3 V.

The integrated circuit DS1305 [21] is a clock/calendar and it also has dual power supply pins for primary and backup power supplies. It is accessed using SPI, and used for time, date, and battery backup management, requesting a processor interrupt to indicate loss of the primary power supply.

The three devices with the highest current consumption are the microcontroller, the LCD, and the DS1305. Their consumption is 30, 2.5, and 1.28 mA, respectively, in the worst case. Memory might need a current peak of up to 25 mA for erase operations and 15 mA for write operations, but in its normal state of stand-by it only needs 1 mA.

I. Physical Features

A photo of an actual system built following the recommendations described in this paper is shown in Fig. 5. As a reference, the size is $38 \times 70 \times 117$ mm and its weight is 265 g, although these data may change depending on the specific components utilized. The current consumption of this recorder is of 42.69 mA (worst case, multiplied by a factor of 1.2), which gives a maximum resulting autonomy of 59 h using two batteries of 1250 mA/h, even though only 48 h are needed. In case the memory were continuously performing erase operations, autonomy would be decreased to 36 h, but this is nonsense in real functioning.

TABLE II
MAIN COMPONENTS LIST

Units	Components	Description
3	INA118	Instrumentation amplifier
4	OPA2336	Double operational amplifier
1	AD7888	Analog to digital converter
1	DS1305	Real time clock
1	T89C51RC2	Microcontroller
1	74AC574	Octal latch
1	PG12864-O	Graphical LCD display
1	MCR102-22RL	Smart media memory connector
1	MAX856	Step-up DC-DC converter
1	MAX749	Fly-back DC-DC converter

III. DISCUSSION

A features comparison between a commercial Holter recorder and ours is shown in Table I. Although our recorder is not intended to be a commercial competitor in the market of Holter recorders, its performance achieves or even surpasses that of a commercial one. Regarding the physical features, they greatly depend on the technical resources available, usually better for a company manufacturing large series of commercial Holter recorders.

This Holter recorder is aimed at filling the gap many research groups have when recording their own signals at a low cost. It can be also very useful as a model to develop other portable signal recorders, since its design is very simple and easily customizable.

It is being used by our research group for signal acquisition. The deparment of cardiology of Alcoy's Hospital is also using it for validation purposes.

IV. CONCLUSION

We described a low-cost microcontroller-based Holter recorder that can be easily implemented with off-the-shelf components (see Table II). The price of all the components needed to build it is around \$150 US. It is intended for research applications that require portable and flexible data acquisition and as a model for other designs. Unlike other similar systems decribed in the literature, the recorder is fully portable and has a performance comparable to that of commercial Holter recorders (see Table I). An example of an ECG obtained with it is shown in Fig. 6.

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